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THE BIOMECHANICAL REFLEXION OF MODERATE IDIOPATHIC SCOLIOSIS IN GAIT CYCLE OF YOUNG ADULTS.

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PREFACE

This study is relative to the three dimensional analysis of the gait cycle in young adults suffering from moderate idiopathic scoliosis (MIS) of the vertebra column. We would like to extract adequate information so as someone to be capable to offer a conservative treatment of gait cycle in close correlation with the human trunk. In the following pages, will would like to present to the reviewer how gait cycle corresponds to a misbalanced posture of the body in scoliosis people and what is the biomechanical reflexion of scoliosis (MIS) according to 3D analysis of gait always compared to healthy people.

CONTENT

CHAPTER 1
THEORITICAL APPROACH OF GAIT

1.1GAIT CYCLE

Gait is defined as the manner or style of walking. One of the attributes of normal walking, as compared with most pathologic gait patterns, is the wide latitude of safe and comfortable walking speeds that are available. Thus, a description of an individual's gait pattern ordinarily includes the speed of locomotion (meters per second) and the number of steps completed per unit of time (steps per minute; this is also called cadence), as well as other characteristics of the gait pattern (Larsson et al, 1980).

During a walking cycle, a given foot is either in contact with the ground (stance phase of the gait cycle) or in the air (swing phase). The duration of the gait cycle for any one limb extends from the time the heel contacts the ground (called heel-strike or heel-on) until the same heel contacts the ground again as illustrated in Figure 1.1. The stance phase begins with initial contact of the foot (usually heel-strike, but in some pathologic conditions, other parts of the foot may contact the ground first) and ends with the foot (usually the ball of the foot and the toes) leaving the ground (called toe-off, or ball-off). The swing phase begins with toe-off and ends with heel-strike. At ordinary walking speeds, the stance phase occupies approximately 60 percent and the swing phase 40 percent of a single gait cycle. Figure 1.1 depicts a full gait cycle for the left and right legs along the same time axis. A typical cycle can be expected to last 1 to 2 seconds, depending on walking speed. Figure 1.1 shows that a period of double support exists when both limbs are in a stance phase. The duration of double support varies inversely with the speed of walking. In slow walking, this period is comparatively long in relation to the swing phase; but as the speed increases, the period becomes shorter and shorter. In running, double support is no longer present. In fact, for a brief time, both feet may be off the ground simultaneously. Each of the two primary phases of the gait cycle can be subdivided into various stages called the sub-phases of gait. For example, the stance phase is comprised of heel-strike, foot-flat, heel-off, and toe-off sub phases.

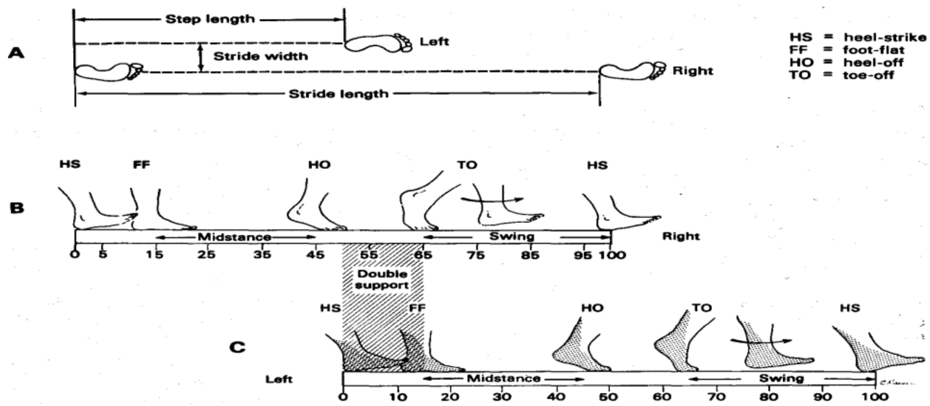


Figure 1.1 Phases of the gait cycle are shown on the same time axis for the left and right leg. (A) Representation of the stride dimensions as viewed from above or beneath the subject. (B) Side view of one complete cycle of the right leg. (C) Side view of one complete cycle of the left leg. The time axes indicate the percentage of the gait cycle completed, starting and ending with heel-strike (HS). Note that two steps occur during each stride. (Adapted from Rosse, C and Clawson, K, 1980.)

1.2PHASES OF THE GAIT CYCLE

In order to provide the basic functions required for walking, each stride involves an ever-changing alignment between the body and the supporting foot during stance and selective advancement of the limb segments in swing. These reactions result in a series of motion patterns performed by the hip, knee and ankle. Early in the development of gait analysis the investigators recognized that each pattern of motion related to a different functional demand and designated them as the phases of gait. Further experience in correlating the data has progressively expanded the number of gait phases identified. It now is evident that each stride contains eight functional patterns. Technically these are sub phases, as the basic divisions of the gait cycle are stance and swing, but common practice also calls the functional intervals phases.

In the past it has been the custom to use normal events as the critical actions separating the phases. While this practice proved appropriate for the amputee, it often failed to accommodate the gait deviations of patients impaired by paralysis or arthritis. For example, the onset of stance customarily has been called heel strike; yet the heel of a paralytic patient may never contact the ground or do so much later in the gait cycle. Similarly initial floor contact may be by the whole foot (flatfoot), rather than having forefoot contact occurs later, after a period of heel-only support. To avoid these difficulties and other areas of confusion, the Rancho Los Amigos gait analysis committee developed a generic terminology for the functional phases of gait. Analysis of a person's walking pattern by phases more directly identifies the functional significance of the different motions occurring at the individual joints. The phases of gait also provide a means for correlating the simultaneous actions of the individual joints into patterns of total limb function. This is a particularly important approach for interpreting the functional effects of disability. The relative significance of one joint's motion compared to the others varies among the gait phases. Also, a posture that is appropriate in one gait phase would signify dysfunction at another point in the stride, because the functional need has changed. As a result, both timing and joint angle are very significant. This latter fact adds to the complexities of gait analysis.

Each of the eight gait phases has a functional objective and a critical pattern of selective synergistic motion to accomplish this goal. The sequential combination of the phases also enables the limb to accomplish three basic tasks. These are weight acceptance (WA), single limb support (SLS) and limb advancement (LA) (Table 1.1). Weight acceptance begins the stance period and uses the first two gait phases (initial contact and loading response). Single limb support continues stance with the next two phases of gait (mid stance and terminal stance). Limb advancement begins in the final phase of stance (pre-swing) and then continues through the three phases of swing (initial swing, midswing and terminal swing).

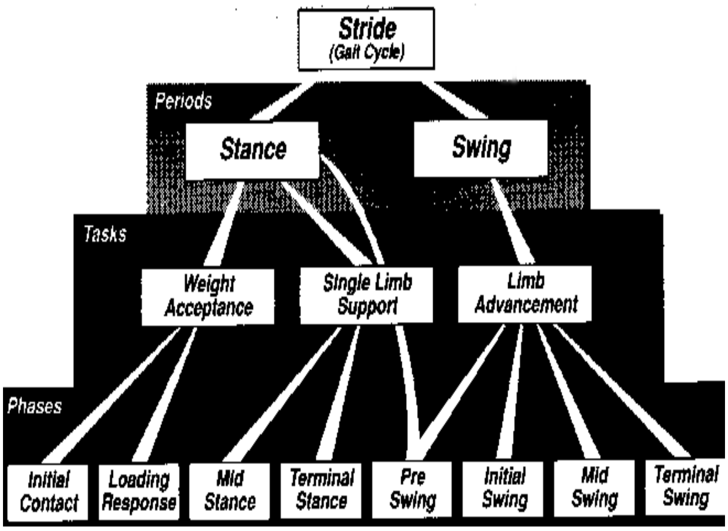


Table 1.1 Divisions of the gait cycle (Adapted from Perry J. MD, 1992)

Task A: Weight Acceptance

This is the most demanding task in the gait cycle. Three functional patterns are needed: shock absorption, initial limb stability and the preservation of progression. The challenge is the abrupt transfer of body weight onto a limb that has just finished swinging forward and has an unstable alignment. Two gait Phases are involved, initial contact and loading response (Table 1.1).

Phase 1-Initial Contact

Interval: 0-2% GC

This phase includes the moment when the foot just touches the floor (Figure 1.2). The joint postures present at this time determine the limb's loading response pattern.

Objective:

- The limb is positioned to start stance with a heel rocker.

Phase 2-Loading Response

Interval: 0-10% GC

This is the initial double stance period (Figure 1.2). The phase begins with initial floor contact and continues until the other foot is lifted for swing.

Objectives:

- Shock absorption
- Weight-bearing stability
- Preservation of progression

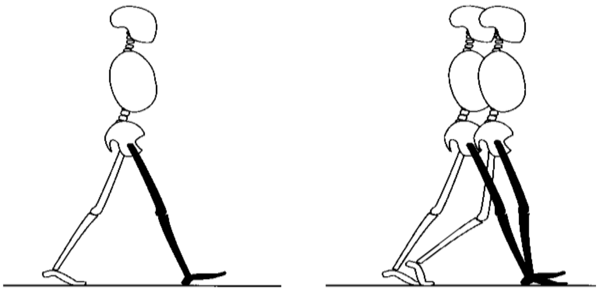


Figure 1.2 Initial Contact: The hip is flexed, the knee is extended, the ankle is dorsiflexed to neutral. Floor contact is made with the heel. Shading indicates the reference limb. The other limb (clear) is at the end of terminal stance. **Loading Response:** The body weight is transferred onto the forward limb (shaded). Using the heel as a rocker, the knee is flexed for shock absorption. Ankle plantar flexion limits the heel rocker by forefoot contact with the floor. The opposite limb (clear) is in its pre-swing phase (Adapted from Perry J. MD, 1992).

Task B: Single Limb Support

Lifting the other foot for swing begins the single limb support interval for the stance limb. This continues until the opposite foot again contacts the floor. During the resulting interval, one limb has the total responsibility for supporting body weight in both the sagittal and coronal planes while progression must be continued. Two phases are involved in single limb support: mid stance and terminal stance. They are differentiated primarily by their mechanisms of progression.

Phase 3-Mid Stance

Interval: 10-30% GC

This is the first half of the single limb support interval (Figure 1.3). It begins as the other foot is lifted and continues until body weight is aligned over the forefoot.

Objectives:

- Progression over the stationary foot
- Limb and trunk stability

Phase 4-Terminal Stance

Interval: 30-50% GC

This phase completes single limb support (Figure 1.3). It begins with heel rise and continues until the other foot strikes the ground. Throughout this phase body weight moves ahead of the forefoot.

Objective:

- Progression of the body beyond the supporting foot

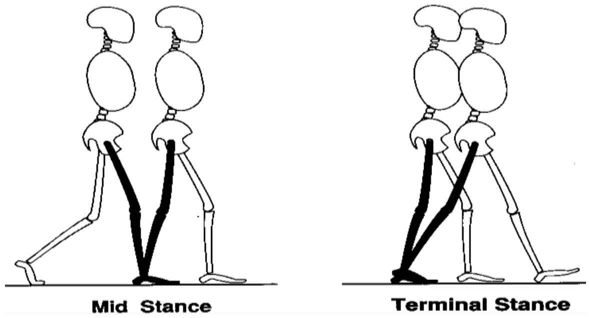


Figure 1.3 Mid Stance: In the first hag of single limb support, the limb (shaded) advances over the stationary foot by ankle dorsiflexion (ankle rocker) while the knee and hip extend. The opposite limb (clear) is advancing in its mid swing phase. **Terminal Stance:** During the second half of single limb support, the heel rises and the limb (shaded) advances over the forefoot rocker. The knee increases its extension and then just begins to flex slightly. Increased hip extension puts the limb in a more trailing position. The other limb (clear) is in terminal swing (Adapted from Perry J. MD, 1992).

Task C: Limb Advancement

To meet the high demands of advancing the limb, preparatory posturing begins in stance. Then the limb swings through three postures as it lifts itself, advances and prepares for the next stance interval. Four gait phases are involved: pre-swing (end of stance), initial swing, mid swing and terminal swing.

Phase 5-Pre-Swing

Interval: 50-60% GC

This final phase of stance is the second (terminal) double stance interval in the gait cycle (Figure 1.4). It begins with initial contact of the opposite limb and ends with ipsilateral toe-off.

Weight release and *weight transfer* are other titles some investigators give to this phase. While the abrupt transfer of body weight promptly unloads the limb, this extremity makes no active contribution to the event. Instead, the unloaded limb uses its freedom to prepare for the rapid demands of swing. All the motions and muscle actions occurring at this time relate to this latter task. Hence, the term pre-swing is more representative of its functional commitment.

Objective:

- Position the limb for swing

Phase 6-Initial Swing

Interval: 60-73% GC

This first phase is approximately one-third of the swing period (Figure 1.4). It begins with lift of the foot from the floor and ends when the swinging foot is opposite the stance foot.

Objectives:

- Foot clearance of the floor
- Advancement of the limb from its trailing position

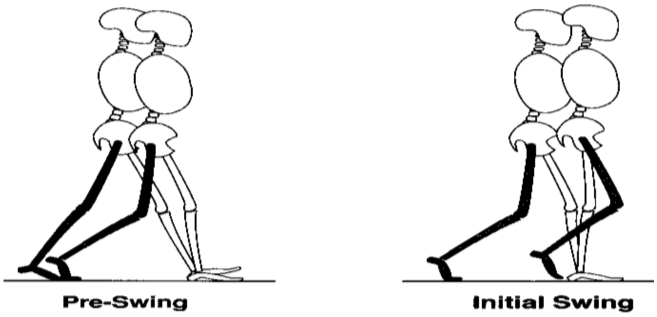


Figure 1.4 Pre-Swing: The Floor contact by the other limb (clear) has started terminal double support. The reference limb (shaded) responds with increased ankle plantar flexion, greater knee flexion and loss of hip extension. The opposite (clear) limb is in Loading Response. **Initial Swing:** The foot is lifted and limb advanced by hip flexion and increased knee flexion. The ankle only partially dorsiflexes. The other limb (clear) is in early mid stance (Adapted from Perry J. MD, 1992).

Phase 7-Mid Swing

Interval: 73-87% GC

This second phase of the swing period begins as the swinging limb is opposite the stance limb (Figure 1.5). The phase ends when the swinging limb is forward and the tibia is vertical (i.e., hip and knee flexion postures are equal).

Objectives:

- Limb advancement
- Foot clearance from the floor

Phase 8-Terminal Swing

Interval: 87-100% GC

This final phase of swing begins with a vertical tibia and ends when the foot strikes the floor (Figure 1.5). Limb advancement is completed as the leg (shank) moves ahead of the thigh.

Objectives:

- Complete limb advancement
- Prepare the limb for stance

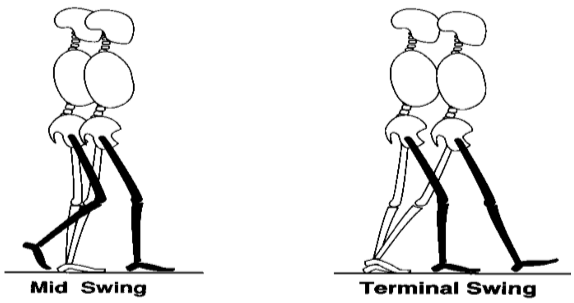


Figure 1.5 Mid Swing: Advancement of the limb (shaded) anterior to the body weight line is gained by further hip flexion. The knee is allowed to extend in response to gravity while the ankle continues dorsiflexing to neutral. The other limb (clear) is in late mid stance. **Terminal Swing:** The limb advancement is completed by knee extension. The hip maintains its earlier flexion, and the ankle remains dorsiflexed to neutral. The other limb (clear) is in terminal stance (Adapted from Perry J. MD, 1992).

1.3WALKING SPEED

Walking speed is an important factor in gait analysis because changes in speed are accompanied by changes in every aspect of walking, including time and distance measurements, energy expenditure, and muscle activity. A simple way to measure average speed, stride length, and cadence is to time the subject walking across a measured distance of at least 15 M and to count the number of steps taken. Normal subjects have the ability to alter their speed of walking from a stroll to a fast walk and into a run, thus making comparisons difficult. Each person, however, has a free or comfortable walking speed on a smooth, level surface that is most energy efficient for that individual. Perry (1992) measured the mean velocity of adults walking a free pace as 82 M per minute, or approximately 3 miles per hour. Their stride length averaged 1.4 M and the mean cadence was 113 steps per minute. Men walked faster and had a longer stride length and a slower cadence than women (Table 1.2). Only a part of the variability of stride is due to leg length. The free walking speed is often used in gait studies because it represents optimal efficiency for each individual, and the able-bodied subject reproduces the same values if the walking surface and the footwear remain the same (Inman, Ralston, and Todd, 1981). The authors classify medium walking speed for men as 100 to 120 steps per minute and for women, 105 to 125 steps per minute. Rates above or below these values are classified as fast or slow walking speeds.

Changes in walking speed are made by altering stride length or cadence, and usually the normal subject changes both parameters. Increased speed results in diminished duration of all of the component phases of the walking cycle (stance, swing, double support) with the double support phase decreasing toward zero and the swing phase decreasing the least. In the running subject, there is no double support period and the swing phase is longer than the stance phase.

	Males	**Females**	**Total**
Number of subjects	135	158	293
Velocity (meters per minute)	86	77	82
Stride length (meters)	1.46	1.28	1.41
Cadence (steps per minute)	111	117	113

Table 1.2 Mean Stride Values in Normal Adults 20 to 80 Years of Age Walking at Free or Customary Walking Speed on a Smooth Level Surface (Adapted from Perry. J, 1992).

1.4PATHOLOGIC MECHANISM OF DEFORMITY

While the long list of diseases that impair patients' ability to walk may differ markedly in their primary pathology, the abnormalities they impose on the mechanics of walking fall into four functional categories. These are deformity, muscle weakness, impaired control and pain. Each category has typical modes of functional impairment. Awareness of these characteristics allows the examiner to better differentiate primary impairment from substitutive actions.

Deformity

A functional deformity exists when the tissues do not allow sufficient passive mobility for the patients to attain the normal postures and ranges of motion used in walking. Contracture is the most common cause. Abnormal joint contours and ankylosis (bony rigidity) also may occur.

A contracture represents structural change within the fibrous connective tissue component of muscles, ligaments or joint capsule following prolonged inactivity or scarring from injury. Relative density, as well as maturity of the connective tissue leads to two clinical contracture patterns: elastic and rigid.

An elastic contracture is one that yields to forceful manual stretch such as using one's whole triceps and shoulder strength (Figure 1.6). The force of two fingers is sufficient to move any normal joint through its full range. Failure to sense the stretch force required causes the examiner to miss the contracture. An elastic contracture presents a confusing picture during walking. In swing the limitations of the contracture will be apparent as the muscles are not programmed to pull harder. Then, during stance, body weight will stretch the tissues so that passive mobility may appear normal or just slightly delayed.

A rigid contracture is one that resists all stretching efforts. Its effect will be consistent throughout the stride. Each joint presents a specific problem.

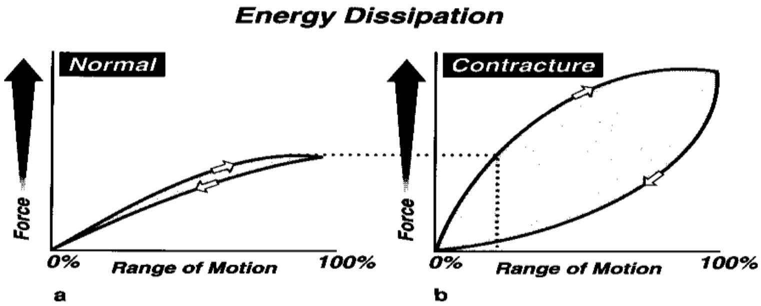


Figure 1.6 Energy absorption by tissues during passive motion. Black lines -force involved. Tissue stiffness indicated by width of space between flexion (up) and extension (down) force curves. (a) Normal tissue flexibility takes minimal energy (b) Contractures absorb greater energy proportional to their tissue stiffness (Adapted from Perry J. MD, 1992).

Muscle Weakness

The patient's problem is insufficient muscle strength to meet the demands of walking. Disuse muscular atrophy as well as neurological impairment may contribute to this limitation. When the cause is a lower motor neuron disease or muscular pathology (poliomyelitis, Guillain-Barre' syndrome, muscular dystrophy, primary muscular atrophy), the patients have an excellent capacity to substitute. With normal sensation and selective neuromuscular control, patients with just muscle weakness can modify the timing of muscle action to avoid threatening postures and induce protective alignment during stance. Similarly, they find subtle ways to advance the limb in swing. Each major muscle group has a postural substitution. Patients also reduce the demand by walking at a slower speed.

If the multiplicity of muscle involvement or a contracture prevents the essential substitutions then the muscles may be suddenly overwhelmed. This is almost an all-or-none situation, that is, either the joints are or are not stable at any one moment. Because the patients do so well when they can substitute, clinicians tend to expect too much from a weakened muscle.

Sensory Loss

Proprioceptive impairment obstructs walking because it prevents the patient from knowing the position of the hip, knee, ankle or foot and the type of contact with the floor. As a result, the patient does not know when it is safe to transfer body weight onto the limb. Persons with intact motor control may substitute by keeping the knee locked or hitting the floor with extra vigor to emphasize the moment of contact. The mixture of sensory impairment and muscle weakness prevents rapid substitution. Hence, even with moderate sensory impairment, walking is slow and cautious. When there is a greater deficit, the patient will be unable to use his available motor control because he cannot trust the motions that occur.

As sensory impairment is not visible, it tends to be ignored. Also, proprioceptive grading is quite crude. There are only three grades: absent, impaired and normal. A grade of normal should not be given unless the responses are rapid as well as consistently correct. Hesitation, as well as the occasional error, is a sign of impairment. A slow response is equivalent to not having sufficient time to catch an overly flexed knee or inverted foot during walking. Consequently, the assessment of proprioception must be critical.

Pain

Excessive tissue tension is the primary cause of musculoskeletal pain. Joint distension related to trauma or arthritis is the common situation. The physiological reactions to pain introduce two obstacles to effective walking: deformity and muscular weakness.

Deformity results from the natural resting postures of a swollen joint. Experimentally, this has been shown to be the position of minimal intra-articular pressure with movement in either direction increasing the joint tension. These minimal intra-articular pressure findings also identify the joint position where the capsule and ligaments are loosest. One can consider these to be the postures that will be assumed by any resting joint.

Muscle weakness occurs secondary to the pain of joint swelling causing reduced activity. Experimental distention of the knee with sterile plasma increased intra-arterial pressure, while quadriceps activation became progressively more difficult. After the pressure prevented all muscle action, anesthetizing the joint restored full quadriceps function (Figure 1.7). This reaction indicated that there is a feedback mechanism designed to protect the joint structures from destructive pressure. Patients display the cumulative effect of this protective reflex as disuse atrophy. During gait analysis, the examiner should expect less available strength and increased protective posturing when the joints are swollen.

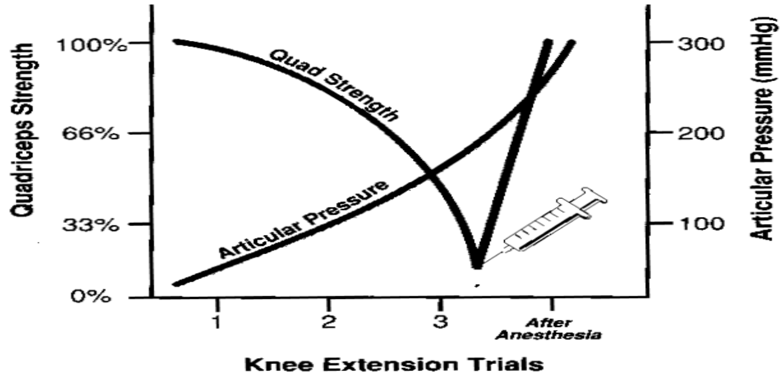


Figure 1.7 Quadriceps inhibition with knee joint distension. Quadriceps strength (top curve) decreases as articular pressure is increased (bottom curve). Injection (syringe) of an anesthetic into swollen joint restores full quadriceps strength (vertical line), (Adapted from Perry J. MD, 1992).

Impaired Motor Control (Spasticity)

Patients with a central neurological lesion (brain or spinal cord) that results in spastic paralysis develop five types of function deficits in varying mixtures and to differing extents. The basic effect is an overreaction to stretch (i.e., spasticity). Selective control is impaired. Primitive locomotor patterns emerge. Muscles change their phasing. Proprioception may be altered. In addition, muscular control is altered by limb position and body alignment. The most common causes of a spastic gait are cerebral palsy, strokes, brain injury, incomplete spinal cord injury and multiple sclerosis.

Spasticity obstructs the yielding quality of eccentric muscle action during stance. The presence of spasticity is readily apparent when a quick stretch induces clonus (Figure 1.8a). Hypersensitivity of the muscles to slow stretch, however, may be missed (Figure 1.8b). Soleus and gastrocnemius spasticity lead to persistent ankle plantar flexion. Progression is obstructed by loss of the ankle rocker and inability to rise on the metatarsal heads for the forefoot rocker. The persistent knee flexion that follows hamstring spasticity limits the effectiveness of terminal swing and restricts thigh advancement in stance. Hip flexor spasticity similarly restricts progression in mid and terminal stance, while sustained quadriceps action inhibits pre-swing preparation for limb advancement.

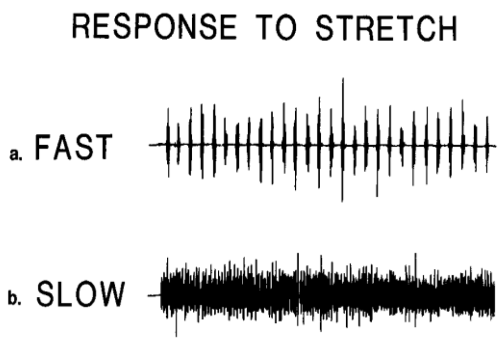


Figure 1.8 Spastic muscle response to stretch (EMG). (a) Fast stretch elicits clonus. (b) Slow stretch generates sustained muscle action (Adapted from Perry J. MD, 1992).

1.5GAIT DEVIATIONS

Hip gait deviations

The multidirectional mobility of the hip makes this joint sensitive to dysfunction in all three planes. A further complexity to assessing the effects of hip pathology is its role as the junction between the lower limb and trunk. Abnormal hip function may be displayed by malalignment of either the thigh or pelvis (and indirectly the trunk). Pelvic motion may accompany the displacement of the thigh, remain stationary, or move in the opposite direction, depending on the mobility of its articulation with the trunk. Thus, in the assessment of walking, thigh motion analysis should be separated from that of the pelvis. The functional patterns of both segments, however, are influenced by the interplay between postural demand and hip joint mechanics (mobility and the actions of its controlling muscles).

The potential gait errors in the sagittal plane include:

- inadequate extension during midstance and terminal stance
- excessive flexion during pre-swing, mid swing and initial swing
- inadequate flexion during initial swing, mid swing, terminal swing, initial contact and loading response

Deviations in the other planes are:

- excessive adduction during the weight bearing period in the stance phase and through the whole swing phase
- abduction that exist during the stance phase and transverse rotation (internal or external)

	Inadequate Extension	Excessive Flexion	Inadequate Flexion	Excessive Extension
Flexion Contracture		X	X	
IT Band Contracture		X	X	
Flexor Spasticity		X	X	
Arthrodesis	X	X	X	X
Pain		X	X	
Voluntary	X	X	X	

	Excessive Adduction	Excessive Abduction	Excessive Rotation
Abduction Weakness	I	C	
Adduction Contracture Spasticity	I	C	
Scoliotic Pelvic Obliquity	I/C	I/C	
Abduction Contracture	C	I	
Iliotiblal Band Contracture	C	I	
Arthrodesis	X	X	X
Voluntary		I	X
Muscle Overactivity			X
Hip Anteversion			X

Key: C = Contralateral
I = Ipsilateral
I/C = Ipsilateral and Contralateral
X = Not side-oriented

Table 1.3 Causes of gait deviations at the hip (Adapted from Perry J. MD, 1992).

Knee gait deviations

The most common types of knee dysfunction occur in the sagittal plane. Inappropriate arcs of motion result in excessive or inadequate flexion or extension. Less frequent are the deviations in the coronal plane (excessive valgus or varus). Excessive transverse plane rotation within the knee is reported, but the findings vary with the method of measurement. This results in a major inconsistency between laboratories, though each facility has confidence in its technique.

	LR	MS	TS	PSw	ISw	MSw	TSw
Inadequate Flexion	x			x	x	x	
Excessive Extension							
Extensor Thrust		x					
Hyperextension		x	x	x			
Excessive Flexion	x			x	x	x	
Inadequate Extension		x	x				x
Coronal Gait Deviations							
Varus	x	x	x				
Valgus	x	x	x				

Table 1.4 Phasing of the gait deviations at the knee (Adapted from Perry J. MD, 1992).

Ankle and foot deviations

Clinical visibility of some gait errors at the ankle has introduced the terms equinus for toe walkers and drop foot for the flaccid foot in swing. They, however, do not cover all the situations involving excessive plantar flexion. Similarly, the opposite term of calcaneus has limitations when one considers all of the possibilities for excessive ankle dorsiflexion. Further confusion in terminology is added by the fact that normal ankle function involves alternate arcs of dorsiflexion and plantar flexion. Consequently, the same gait error could be classified as excessive plantar flexion or inadequate dorsiflexion. This is particularly true for those phases where neutral ankle alignment is expected and the common functional error is failure to fully dorsiflex the foot from a previous planter flexed posture. Conversely, during those gait phases where some degree of plantar flexion is normal, failure to attain the appropriate arc can be either inadequate plantarflexion or excessive dorsiflexion.

A preliminary trial with such duplicative terminology proved very confusing. As a result, all the functional errors at the ankle will be classified as either excessive plantarflexion or excessive dorsiflexion.

Excessive ankle plantar flexion occurs at (Perry, 1992):

- Initial contact and Loading response phases
- Mid and terminal stance
- Pre-swing, initial and terminal swing

Excessive ankle dorsiflexion occurs at:

- Initial contact and Loading response phases
- Mid and terminal stance
- Pre-swing, initial, mid and terminal swing

Abnormal function of the foot during walking may be displayed by two situations. These are the pattern of foot contact during the stance phase and malalignment of the foot in swing. The cause may be a reflection of knee and dysfunction or intrinsic foot pathology. Deviations occur in both the sagittal and coronal planes.

CHAPTER 2
THEORITICAL APPROACH OF SCOLIOSIS

2.1BIOMECHANICAL DEFINITION OF SCOLIOSIS

Scoliosis is defined as an appreciable lateral deviation in the normally straight vertical line of the spine. Since the ultimate effect of the disease is an extensive alteration in the mechanical structure of the spine, a biomechanical definition of the disease is necessary. There is abnormal deformation between and within vertebrae, too much curvature in the frontal plane, too much vertical axis rotation in the wrong direction, and not enough curvature in the sagittal plane (a loss of normal kyphosis or lordosis). It should be emphasized that there is also deformity that may not be recognized in the analysis of the traditional orthogonal planes.

In other words, the relative position of vertebrae in regions of the spinal column is abnormal, and deformation within an individual vertebra is abnormal. There is too much curvature. Instead straight spine in the frontal plane (x, y plane) or the subtle, right physiologic curve, there is an exaggerated curvature in the frontal plane. The curves are in the wrong plane. Generous curves in the sagittal plane are normal. (There is normal cervical and lumbar lordosis and thoracic kyphosis.) The axial rotation is in a direction opposite of what would be expected from the physiologic coupling between lateral bending and axial rotation. In scoliosis (Figure 2.1), there is considerable deformation within a given vertebra. There may be wide pedicles on one side and a short pedicle on the other. The transverse processes may be asymmetrical in their spatial orientation. The spinous process may be deformed and bent out of the midline. The laminae and the vertebral bodies are asymmetrical (White A. and Panjabi M., 1990).

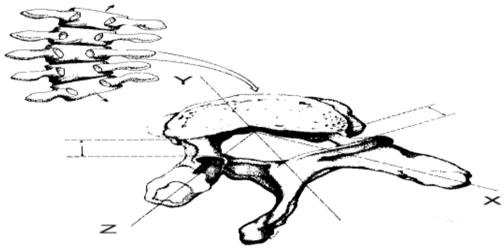


Figure 2.1 This diagram emphasizes that in scoliosis there is deformity within as well as among a vertebrae. Studies by Sevastik, et al., 1984, show that deformation within the vertebra does not occur in curves with Cobb angles of less than 40^0. Note the abnormal configuration and spatial orientation of the pedicles. Consider the potential value of studying this in the preoperative preparation and planning activities in which transpedicular fixation is to be employed (Adapted from White A. and Panjabi M., 1990).

2.2ETIOLOGIC CONSIDERATIONS

Biomechanic Classification

There is a long list of known causes, conditions, and diseases that are associated with scoliosis. There are several methods of classification. A biomechanical classification is provided here and may best be appreciated in the following context. The spine remains normal because of the maintenance of a delicate and precarious balance. This balance depends on a precise functional status and dynamic symmetry, the key elements being the bony structure, the ligaments, the intrinsic neuromuscular mechanics, and finally, the general balance and symmetry of the body. Scoliosis can result from either gross or subtle disruptions of the delicate balance. The diseases listed are not exhaustive for each category. There is also some overlap; a given disease may contribute or be presumed to contribute to the imbalance through more than one mechanism.

BIOMECHANICAL CLASSIFICATION OF SCOLIOSIS

Alterations of Intrinsic Osseous Structures

Abnormalities of Material Properties of Support Structure

Rickets (primary and secondary)   Neurofibromatosis

Osteogenesis imperfecta   Infections or tumors

Abnormalities of the Geometry of the Support Structure

Hemivertebrae

Maldeveloped vertebrae

Myelomeningocele

Asymmetrical spina bifida

Asymmetrical lumbosacral vertebral structure and articulation

Fractures and dislocations

Various surgical procedures

Postirradiation (vertebral end-plates)

Abnormal Regional Kinematics

Congenital unilateral bars

Partial failures of segmentation

Asymmetrical sacralization of fifth lumbar vertebra

Fractures, dislocations and surgery

Alterations of Intrinsic Ligamentous Structures

Marfan's disease

Mucopolysaccharidosis

Myelomeningocele

Arthrogryposis

Surgery

Alterations in Static or Dynamic Balance

Neuromuscular Static Balance

Polio

Myelomeningocele

Syringomyelia

Neuromuscular Dynamic Balance

Cerebral palsy

Friedreich's ataxia

Muscular dystrophy

Postural Dynamic Balance

Abnormalities of vestibular apparatus

Visual disturbances

Torticollis

Leg-length discrepancies

Thoracic Static Balance

Rib removal (thoracoplasty); ipsilateral convexity

Excessive thoracic scarring; contralateral convexity

Congenital Scoliosis (Deformity Intrinsic to Body)

Infantile type

Sprengel's deformity

Klippel-Feil syndrome

Multiple congenital anomalies

Idiopathic Scoliosis

Syndrome of contractures

Miscellaneous Forms of Scoliosis

Experimental and Clinical Studies

The cause of 65% of cases of scoliosis is unknown. Idiopathic scoliosis occurs in an otherwise healthy child, often associated with a familial history of the disease. Numerous hypothetical, etiologic explanations are offered. From a mechanical point of view, the hypothesis should explain the cause of the abnormal curvatures, the abnormal rotation, and the forces necessary to cause deformation within a vertebra.

One of the major experimental thrusts has been to establish some imbalance in the neuromuscular and osseous ligamentous structures of the spine in experimental animals, the assumption being that imbalances that result in a scoliotic pattern may be sought as potential etiologic factors in idiopathic scoliosis. This presumes that scoliosis is caused by the weakness or absence of a structure on the convex side of the curve or an overactivity of its antagonist on the concave side. A large number of anatomic elements have been studied in rabbits and pigs.

When one finds a consistent unilateral alteration resulting in scoliosis, the assumption is that with a growing spine, the initial, functional scoliosis ultimately develops into a structural deformity. This is explained on the basis of what has been traditionally called Heuter-Volkmann's` law. The theory suggests that increased pressure across an epiphyseal growth plate inhibits growth, whereas decreased pressure across the plate tends to accelerate growth. This theory purports that, on the concave side of the curve, the epiphyseal plates have abnormally high pressures that result in decreased growth, whereas on the convex side of the curve the pressures are less, resulting in accelerated growth. These two factors contribute significantly to vertebral asymmetry. Work by Stillwell (1962) on monkeys nicely supports this hypothesis.

Two experiments involved the fixation of the spine in a curved position and fixation of the spinous processes of the vertebrae. The first resulted in occasional scoliosis, and the second resulted in severe scoliosis with lordosis and rotation.

Another etiologic consideration related to mechanics is asymmetrical radiation of the spine, resulting in curvature due to changes in the epiphyseal growth plates, with either unilateral stimulation or unilateral ablation.

Experimental scoliosis has also been produced by radiologic exposure of the growing spine, induction of lathyrism, oxygen deficiency, unilateral labyrinthine stimulation, and unilateral labyrinthine ablation. The very broad variety of experimental variables that have resulted in a "scoliotic" deformity suggests that the maintenance of a normal spine in a growing animal is dependent upon a delicately balanced, easily disrupted equilibrium.

This general experimental approach is questionable because the goal is to explain idiopathic scoliosis in man, an erect, biped organism. The frequency duration, direction, and magnitude of the loads are significantly different in the pig, dog, rabbit, and mouse. There are obvious differences in the anatomy of the spine in animals. Subtle anatomic factors such as facet orientation can significantly alter the mechanics of the spine. Thus, there is yet another limiting factor in the use of animals to study scoliosis in man. Using quadrupeds as experimental prototypes is, no doubt, valuable but should be viewed in this perspective.

Experimental Studies Update

This section includes the views from Augustus White and Manohar Panjabi (1990) relative to some of the more cogent experiments that relate to the etiology of scoliosis.

Lawton and Dickson (1986) have studied New Zealand white rabbits to develop an experimental scoliosis. They created in these animals' pure frontal plane deformities, pure sagittal plane deformities, and a combined sagittal and frontal plane deformity that was called biplanar. They noted that neither pure scoliosis (frontal plane deformity) nor pure lordosis (sagittal plane deformity) resulted in progressive scoliosis. However, all 20 animals given the combined deformity involving both planes developed scoliosis. Moreover, they noted that if the two-plane deformity was released before maturity, the deformity spontaneously improved. The investigators state that this experiment supports the view that the etiology of idiopathic scoliosis is the anterior elements growing faster than the posterior elements, causing a loss of normal kyphosis and a buckling of the anterior elements (vertebral bodies) outward laterally. This creates the scoliosis. The investigators purport that partial correction of the frontal plane deformity alone does not address the important loss of normal kyphosis (lordosis deformity). This is an important clinical biomechanical consideration that requires additional thought and investigation.

In some support of the work of Lawton and Dickson and the hypothesis of Roaf (1966) is the work of Ohlen (1988) and associates. They studied 127 patients with idiopathic scoliosis and noted that they did have less thoracic kyphosis than normal.

Hakkarainen (1981) studied 253 growing rabbits in an experimental design that involved immobilizing the animals in a cast that produced a scoliotic curve of the spine. The animals were left in the cast for 2-5 weeks. The results were that 85% of the animals became scoliotic, and 52% of those animals had either permanent or progressed scoliosis. The curves that were less than 30^0 at the time of cast removal did not progress. On the concave side of the curve there were shortened intercostal muscles and evidence of growth disturbance.

These two studies show that a variety of changes in the normal balanced mechanical relationships can result in either temporary or permanent deformity. Attempts to compensate for or re-establish that delicate balance are sometimes successful and sometimes not. The type, degree, extent, and duration of the imbalance are also factors that influence the probability of return to a normal or new balance, either spontaneously or as a result of some intervention.

There has been considerable interest in a variety of neurophysiologic abnormalities associated with scoliosis. These have been comprehensively summarized by Yamada (1984) and colleagues. The work of Pincott and Tafts (1982) is very interesting because, unlike most animal studies, theirs was done with primates. This experimental scoliosis was actually an incidental finding. Monkeys were injected intraspinally with live attenuated oral poliomyelitis vaccines. Some of the animals developed scoliosis. These animals were found to have damage on the convex side of the spinal cord, particularly in the posterior horn and posterior central gray matter (Clarke's column). The scoliosis was thought not to be due to poliomyelitis, because the anterior horn was not involved. Thus, these data are interpreted as supportive of the theory that asymmetrical weakness of the paraspinal muscles can be due to loss of proprioceptive innervation.

Another relatively rare study in primates by Pincott and associates has shown that resection of the dorsal spinal nerves can create a scoliosis convex to the side of the resection. This supports the hypothesis that scoliosis can be created by asymmetrical spinal muscle weakness due to loss of proprioception.

Yamada and co-workers emphasized that virtually any disruption of the postural reflex system can result in scoliosis. The imbalance may be in the afferent system either primarily or secondarily, as suggested by the preceding experiment. Also, the disruption may be due to disruption of the afferent system. The authors also indicated that there is clinical and experimental evidence that brain stem dysfunction may contribute to the etiology of scoliosis.

These studies show that (1) one can produce scoliosis through the creation of pure sagittal plane lordosis, (2) scoliosis can also be produced through the production of a frontal plane deformity, (3) subtle muscle imbalance secondary to the alteration of sensory input can create scoliosis, and (4) disruptions of postural reflexes probably can produce scoliosis. At the time of this writing, the experimental data support the trend of many possible causes of "idiopathic" scoliosis rather than one common cause.

Clinical Studies Update

The work of Yamada and colleagues summarizes well the clinical evidence for disruption of postural reflex as a cause of scoliosis. The publication by Sevastik and associates (1984) provides a thorough review of the literature. This work emphasizes two factors that may subtly alter the delicate balance that can yield to what sometimes seems like a propensity to develop scoliosis. The asymmetric growth of the ribs (i.e., increased longitudinal growth on the concave side) may be the cause of right convex thoracic scoliosis. It was also suggested that the more pronounced vascularization of the often larger right breast may stimulate enough growth of the underlying right costal cartilage to upset the delicate balance of forces acting on the normal spine. This study also noted with computerized tomography (CT) studies that with Cobb angles of $<40^0$ there was no evidence of asymmetry of vertebral bodies, pedicles, laminae, or transverse processes in transverse sections of the *apical* vertebra.

Yarom and Robin (1979), in a study of spinal and peripheral muscles in patients, noted some abnormalities in the concave side of the curve. They reported a mild Type I fiber atrophy and a generalized tendency toward small myofibrils. This suggests a generalized muscle disorder with asymmetrical changes upsetting the delicate balance.

A not-so-subtle form of imbalance is the scoliosis seen in myelodysplasia. Here, there is gross muscle imbalance and paralysis, as well as incomplete and asymmetrical posterior element structural imbalance.

Another not-so-delicate disequilibrium of the delicate balance of a normal growing spine is created by chest wall resections in children. Derosa noted that children with posterior, not anterior, chest wall resection developed progressive curvature with the convexity always to the normal side. This clinical study fits neatly with the animal investigations of Langenskibld and Michelsson (1962), who found that among the 15 operations that create experimental scoliosis, the one best able to do so was paravertebral rib resection.

Mayfield and associates(1981) have completed clinical studies that show another mechanism for the development of scoliosis. Children treated with orthovoltage radiation (>3,000 rads) for neuroblastoma may develop scoliosis or kyphosis from various patterns of location of epiphyseal arrests. This can result in a variety of patterns of progressive deformities.

The issue of leg length difference as an etiologic factor in scoliosis has been studied (Papaioarmou et al, 1982). The associated lumbar scoliosis was observed in a study of 23 young adults to be compensatory and nonprogressive. The scoliosis was minor in patient with less than 2.2-cm discrepancies. In a study of 15 patients with leg length differences averaging 3 cm (1.4-5.5 cm), the development of scoliosis was noted. Neither of these studies noted significant back pain. Both reported some residual scoliosis following correction of leg length discrepancy with lifts.

These updated clinical studies support the observations from experimental studies that disruption of postural reflexes, both subtle and dramatic muscle imbalance, resection of the posterior portion of the thoracic cage, radiation therapy, and leg length discrepancies are all capable of producing or causing scoliosis.

2.3ETIOLOGIC THEORIES

A review of the salient theories that have a basis in biomechanical principles follows.

Roaf (1966) suggests that the basic problem in scoliosis is relative lengthening of the anterior components of the spine compared to the posterior elements. Such a situation in an unyielding anterior musculoskeletal wall of the body results in lateral deviation of the spine and the subsequent development of scoliosis. The theory does not explain why the deviation is so predominantly to the right. Also, there is unconvincing evidence that the muscles of the anterior abdominal wall cannot stretch, yield, or accommodate the long anterior elements. Moreover, the muscles are not particularly tense in patients with scoliosis.

MacEwen (1968) produced scoliosis experimentally in animals by transection of the dorsal nerve root and suggested that the result may be due to a loss of sensory input. The convexity of the resultant curve was to the side of the disrupted neural sensory elements. Alexander, Bunch, and Ebbesson (1973) showed with histologic staining techniques and examination of the anterior nerve cells that the ablation of the dorsal sensory roots also caused an associated motor impairment.

A theory proposed by White (1971) is as follows. The observation that occasional coupling of axial rotations of the vertebrae causes the anterior aspect to point toward the convexity of the lateral curve in normal bending in the middle portion of the thoracic spine must be considered. It is generally acknowledged that scoliosis frequently starts in this midthoracic area. There is already a physiologic, slight, right thoracic curve in this region. If some precarious balance of the normal thoracic motion should be disturbed, vertebrae in the physiologic, right thoracic curve might somehow rotate too much into the convexity of the curve (Figure 2.2). Such an occurrence could set off a chain of events leading to asymmetrical loads on the epiphyseal plates and muscle and ligamentous imbalance, with ultimate progression to scoliosis. The precipitating condition may be an abnormal or malaligned facet, a discrete traumatic episode, a chemical hormonal change, extreme handedness of the individual, or any number of other possible embarrassments that upset the delicate balance. The crucial variable may be whether or not the thoracic vertebrae of the normal curve rotate toward the concavity or the convexity of the lateral curve.

The relation of handedness to scoliosis has been supported by the observation of the left convex curvature in left-handed individuals. It is tempting to speculate about the great proportion of right thoracic curves in idiopathic scoliosis. Do left-handed individuals with scoliosis tend to have left thoracic curve deformities? McCarver and colleagues reviewed left thoracic and related curve patterns. Of 14 patients with left thoracic curves, ten were right-handed, only two were left-handed, and two were ambidextrous. One left-handed patient and one of the two ambidextrous patients had infantile idiopathic scoliosis. These data do not support a hypothesis suggesting some association of left thoracic deformity with left-handedness.

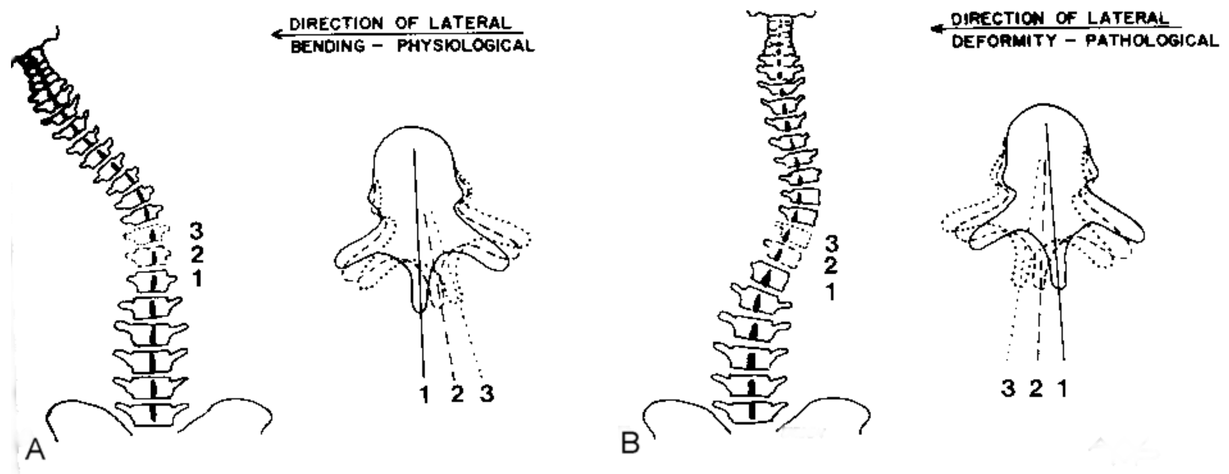


Figure 2.2 Axial rotation into the normal (physiologic) lateral curve (A) and the scoliotic curve (B) is represented. (A) The normal curve generally shows axial rotation of the anterior portion of the vertebra into the concavity of the physiologic curve. However, sometimes the curve may rotate into the convexity. (B) In the scoliotic spine, the associated axial rotation is always into the convexity of the lateral curve. (Adapted from White A. and Panjabi M., 1990).

Another possible neurologic basis for the etiology of scoliosis has been proposed. This is based on epidemiologic studies carried out by Yamada and colleagues. These investigators found that of 100 patients with scoliosis, 99 had abnormal equilibrium. This malfunction progressed with the severity of the scoliotic curve. At full growth, the findings gradually diminished and disappeared. The dysfunction was noted in the proprioceptive and optic reflex systems. This observation shows an association, but not necessarily a cause. Nevertheless, these neurologically based clinical studies provide important information to help understand scoliosis.

Loynes (1972) carefully reviewed 241 patients who had thoracoplasty and removal of three to ten ribs. A convex scoliotic curve to the side of the operation developed in 99% of these patients. Scoliosis tended to progress with time.

Ponseti (1973) suggested that a shift in the position of the nucleus pulposus toward the convex side of the curve might be the cause of scoliosis. The normal physiologic shift of the nucleus pulposus is toward the concavity of the curve.

Occasionally a patient is seen who has a single or double curve over a tilted, asymmetrical, or malformed fifth lumbar vertebra. The situation can also exist with pelvic obliquity. Such asymmetry may result in a moment about the z-axis that would tilt the entire spine off to one side. In order to keep the center of gravity over the sacrum, a physiologic curve develops in the lumbar spine. With time, the unbalanced forces acting on the epiphyses lead to structural changes. If compensation is then needed above the lumbar curve, and the epiphyses are young enough to respond, a similar process may occur in the thoracic curve. That is, a functional curve that was initially compensatory may become structural (Figure 2.3). Our update of the experimental and clinical studies of scoliosis provided no salient new hypotheses regarding the etiology of scoliosis. We believe that the normal spine in a growing person has a precise, precarious, delicate mechanical balance. Asymmetrical changes in primary structures, support structures, growth centers, the position of the spine, and related neural or muscular components can result in the development of scoliosis.

Karski (2002) suggested a biomechanical consideration that connected with the hip and pelvic regions about the Etiology of the so-called "idiopathic scoliosis". At all children with idiopathic scoliosis there is a real or functional abduction contracture of the right hip (sometimes plus flexions- and out-rotation contracture). The right hip abduction contracture is connected with "syndrome of contractures" at newborns and babies. There are two groups of development of idiopathic scoliosis. The first group - small children, early rotation deformity, both scoliosis (L and Th), progression. The second group is connected only with the habit of permanent standing "at ease" on the right leg.

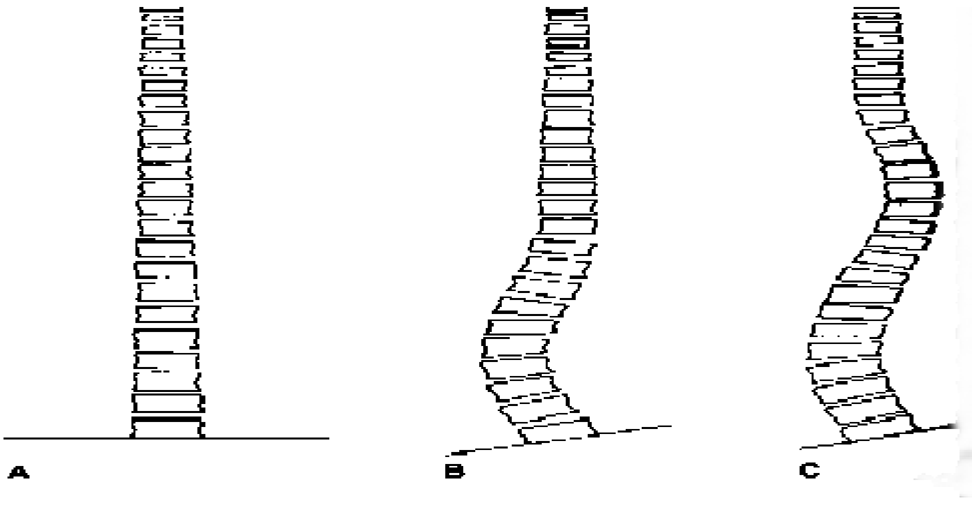


Figure 2.3 (A) A normal balanced spine. (B) "Tilt" or asymmetry in lumbosacral area, with a functional or structural lumbar curve. (C) A superimposed functional and structural thoracic curve (Adapted from White A. and Panjabi M., 1990).

Scoliotic curve patterns classification

The curve patterns of scoliosis, relative to the point where they presented on the spinal column, classified in different groups. Every curve pattern has a different prognosis and so the localization and the classification are of great importance.

The classification is as follows:

- **Thoracic curve pattern.** It exists in 25% of all cases and the curve is towards the right side. The upper vertebra of the curve is $Th_{4\text{-}6}$, while the lower is $Th_{8\text{-}9}$.
- **Thoracolumbar curve pattern.** It exists in 19% of all cases and the curve is towards the right side. The upper vertebra of the curve is $Th_{5\text{-}7}$, while the lower is $L_{1\text{-}4}$. The apex of the curve is at the level of $Th_{11\text{-}12}$.
- **Lumbar curve pattern.** It exists in 25% of all cases and the curve is usually towards the left side. The upper vertebra of the curve is at the level of $Th_{11\text{-}12}$ while the lower is at the level of $L_{3\text{-}5}$. The apex of the curve is at the level of $L_{1\text{-}2}$.
- **Cervicothoracic curve pattern.** The curve is usually towards the left side and is very rare.
- **Double major organic curve pattern.** It exists in 30% of all cases. Such curve pattern is the thoracic, lumbar, thoracolumbar, double thoracic and the low lumbar.

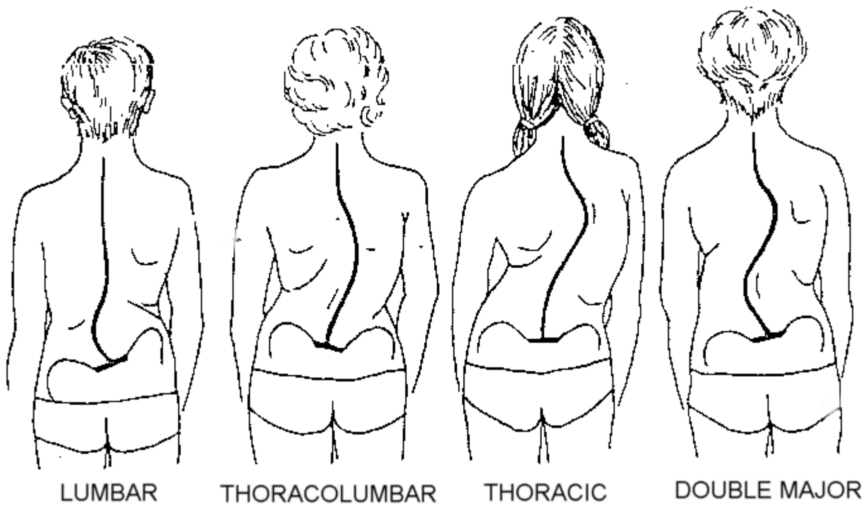


Figure 2.4 Different types of curve patterns. The position and the type of the scoliotic curve pattern may influence the balance and the equilibrium of the whole vertebra column (Adapted from White A. and Panjabi M., 1990).

CHAPTER 3
KINEMATICS THEORY AND JOINT KINEMATICS (CAMERA BASED SYSTEMS)

Human motion (kinematic) studies have been performed for some time in a variety of improvised ways, although the objective was not always a kinetic one. In 1887, Eadweard Muybridge published a set of sequential photographs of the human figure in action performing different activities. This was probably the first artistic attempt to resolve human motion into its kinematic composition. Evidence of attempts to study human gait as a diagnostic tool can be found in the late part of the 19th century. Braune and Fisher recognized the need to know about the mass properties of the human body and were among the first known to have performed a mass distribution study on cadavers. Greater interest in the subject began in the middle of the twentieth century at the outset of the Second World War and continues to the present.

Although the drive for studying locomotion originated from clinical demand, the bulk of the studies focused on the improvement of existing methodology rather than development of clinically viable methods. Some of the most notable early contributions to the study of locomotion are the works by Elftman, Inman, and Brestler.

The latter recognized the need to develop scientific methods of monitoring locomotion. As a result, some kinetic attitudes towards gait analysis evolved, and these remain almost unchanged to a present day. The recognition that motion forces and muscle activity are the fundamental variables in monitoring locomotion is traced back to these early studies. With the advent of modern technology, these techniques have been refined and somewhat adapted for clinical use. Studies were further expanded to cover human performance in a more general sense such as in athletics, rehabilitation, orthopedics, and vocational task performance (Craik R. and Oatis C., 1995).

3.1RIGID BODY MODEL OF THE LOWER LIMB

The structure of a human body is extremely complicated, yet well organized. Basically, it is articulated with hundreds of bones that are covered by soft tissue consisting of muscles, tendons, ligaments, and skin. Such structural complexity makes the exploration of human movement a great challenge.

Fortunately, the primary properties of gross body movement, such as gait, are represented by the bones, although soft tissues undergo various kinds of motions such as stretch, compression, or contraction. Therefore, in this regard, the soft tissues can be ignored and the human body can be considered as a structure that consists only of bones. It should be stressed that soft tissue movement is an important issue in some aspects of gait analysis. However, it is beyond the focus of this book.

Let us consider about a skeletal body. Anatomically, it is still complicated by the facts that (1) every bone has its unique shape, and (2) two adjacent bones are mainly connected through a complicated arrangement of ligaments. Functionally, however-kinematics in particular-it is quite simple. That is, the primary function of the bones is for the transformation of movement from one point of the body to the other, and the function of the connections between the bones is to form a center of rotation so that the two coupled bones can undergo relative rotational movement. Clearly, such functions can be easily replaced by a mechanical structure. For example, a bone could be represented conceptually, without loosing generality, by a rigid mechanical structure, such as a nondeformable cylinder that has identical inertial properties to the bone (see Figure3.1), and the articulations could be represented by mechanical joints, such as hinges, ball and socket joints, or universal joints (see Figure 3.1).

Thus, a sophisticated human skeletal structure can be superseded by a simple mechanical system that consists of multiple rigid segments, or links, and joints. Such a mechanical system is called the rigid body model of the human body.

For various reasons, a rigid body model of the human body could include different numbers of rigid segments. As an example, a fifteen-segment model is illustrated in Figure 3.2. It includes fifteen links: two feet, two shanks, two thighs, one pelvis, one trunk, two upper arms, two forearms, two hands, and one head; and fourteen joints: two ankles, two knees, two hips, one spinal disc, two shoulders, two elbows, two wrists, and one neck.

The model can have more segments by adding flexibility to the foot and the trunk, or have fewer segments by restricting joints such as the ankle or the knee in the lower limb, the elbow or the wrist in the arm, or the spinal disc and the neck in the upper body.

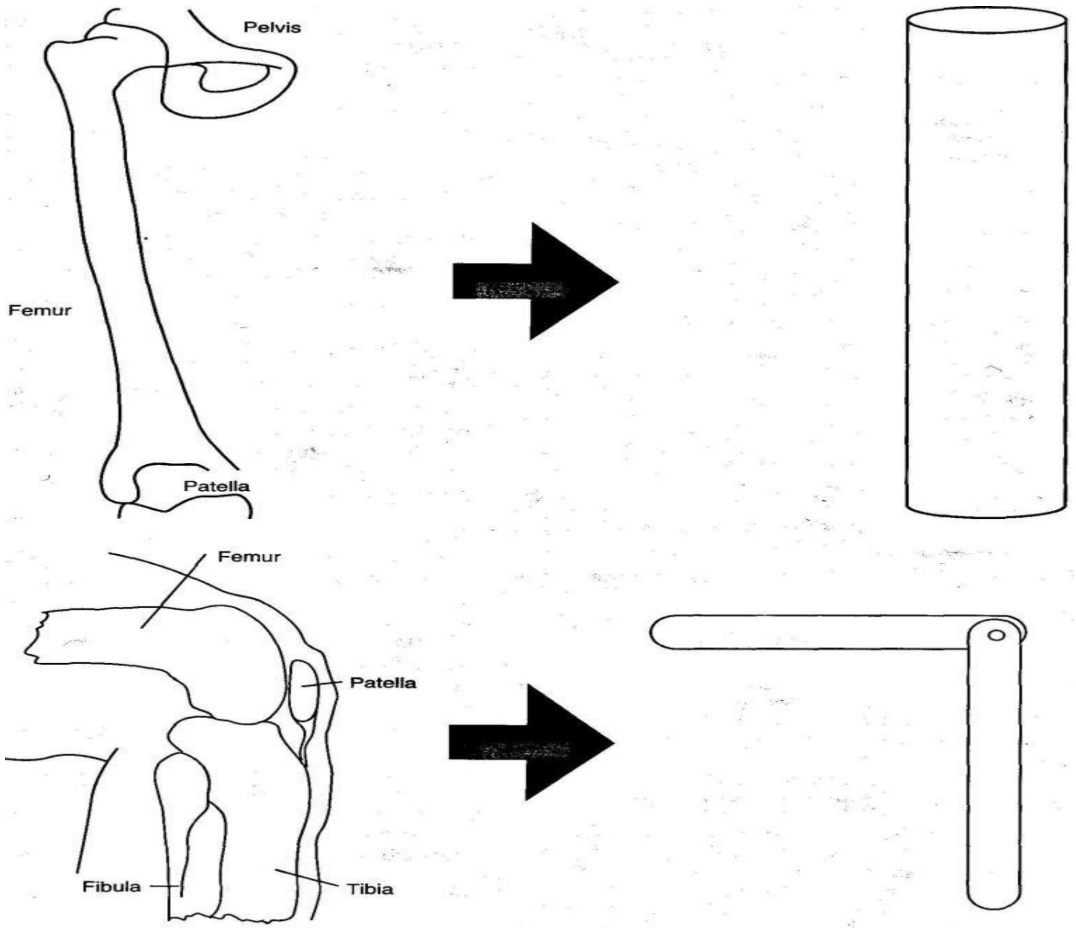


Figure 3.l A femur and a rigid cylinder and a knee joint and its mechanical model, a hinge joint (Adapted from Craik R. and Oatis C., 1995)

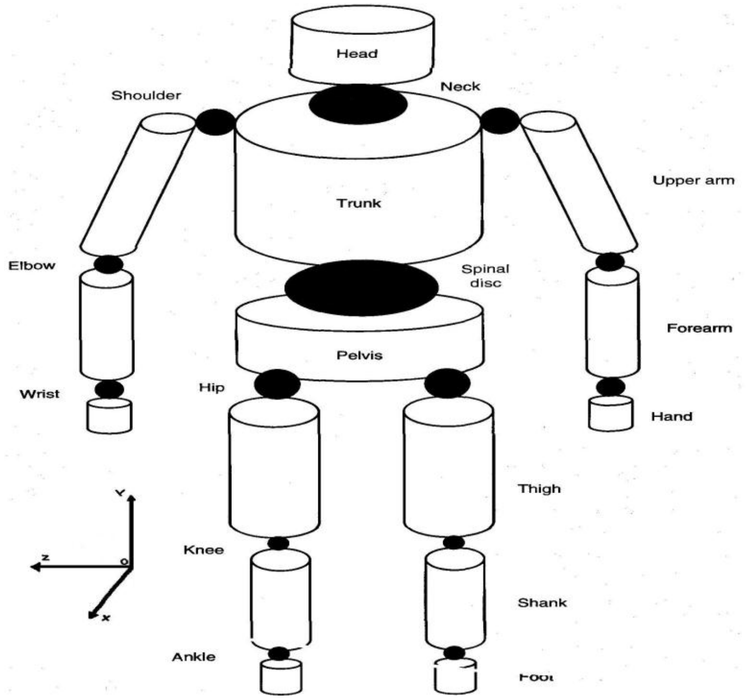


Figure 3.2 A 15 segment rigid body model that includes feet, shanks, thighs, pelvis, trunk, forearms, upper arms, hands and head (Adapted and reprinted from Craik R. and Oatis C., 1995).

3.2REFERENCE SYSTEMS

In order to describe the movement of body segments, it is essential to establish a reference space so that all the description is referred to that space. In general, the reference space is represented by a set of three orthogonal unit vectors, called the Cartesian coordinate system, which is rigidly attached to the space. For gait analysis there are commonly four reference systems: Absolute Reference System, Local Reference System, Segmental Reference System, and Joint Reference System. In the following sections, the definition as well as the determination of these systems will be discussed. It should be noted that no unique, consistent definitions presently exist for these systems.

The ones introduced in this chapter are based on the "Recommendations for Standardization in the Reporting of Kinematics Data" proposed by the International Society of Biomechanics.

Absolute Reference System

The Absolute Reference System is also called the Inertial Reference System or Global Reference System.

Definition: A right-handed orthogonal triad <X, Y, Z > fixed in the ground. While a person is standing in an anatomically defined neutral posture, each of the axes is defined as:

+X axis pointing anteriorly (forward)

+Y axis pointing superiorly (upward) and in parallel with the field of gravity

+Z axis pointing rightward

The Absolute Reference System is depicted in Figure 3.3. The directions of the X, Y and Z axes are chosen so that for those conducting two dimensional studies, X, Y will lie in a sagittal plane. This will be consistent with the three dimensional convention.

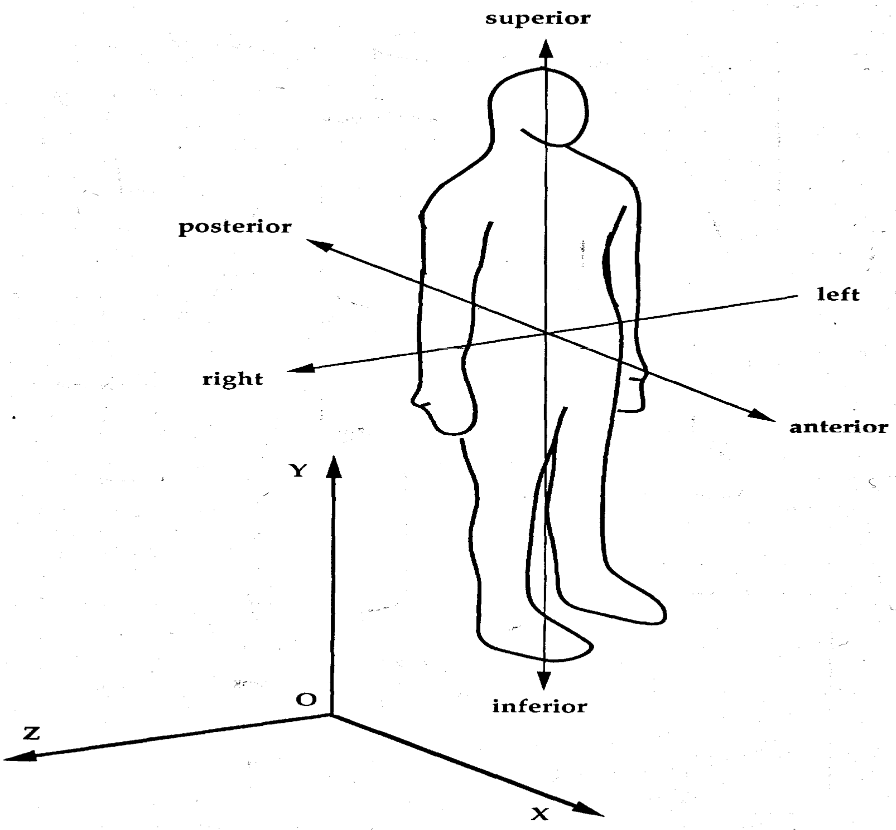


Figure 3.3 Definition of Absolute Reference System (Adapted from Craik R. and Oatis C., 1995).

Local Reference System

The Local Reference System is also called the Body-Fixed Reference System. It is a Cartesian coordinate system that is fixed to and moves with the body. The three axes in this system may or may not have anatomical meanings. Such a system is needed to describe segmental position and orientation with respect to the Absolute Reference System.

Definition: A right-handed orthogonal triad $<X_{Li}, Y_{Li}, Z_{Li}>$ fixed to a point on the i^{th} body segment. Although it may seem trivial to define the exact directions of the axes, we have defined the exact directions of all the axes to keep them consistent throughout the chapter. Thus, the directions of the axes are recommended as:

+ X_{Li} axis pointing forward / backward (sagittal plane)

+ Y_{Li} axis pointing upward / downward (vertical-gravitational plane)

+ Z_{Li} axis pointing rightward (frontal plane-medial / lateral direction)

Obviously, the anatomical directions used in the preceding definition refer to the individual body segment, rather than the whole body. Usually, the Local Reference System is established by a group of markers that are attached to the body. They can be attached directly to the body at various locations, or connected to each other to form a rigid segment and then attached to the body segment as a whole. This latter method is called rigid-segment approach.

Since the former method is the same as the anatomical landmark approach, only the rigid-segment approach is described here. Figure 3.4 illustrates a rigid segment i that includes three markers (1, 2, and 3). An

Absolute Reference System <**X, Y, Z** > is first established so that the coordinates of these three markers are known. That is:

$$\boldsymbol{R}^{1} = [x^{1}\ y^{1}\ z^{1}]^{T} \quad (1)$$

$$\boldsymbol{R}^{2} = [x^{2}\ y^{2}\ z^{2}]^{T} \quad (2)$$

$$\boldsymbol{R}^{3} = [x^{3}\ y^{3}\ z^{3}]^{T} \quad (3)$$

where T denotes the transport operator to the matrix. Based on these three position vectors the Local Reference System $<X_{Li}, Y_{Li}, Z_{Li}>$ can be determines as:

$$\boldsymbol{R}^{1,2} = \boldsymbol{R}^{1} - \boldsymbol{R}^{2} \quad (4) \qquad , \qquad \boldsymbol{R}^{1,3} = \boldsymbol{R}^{1} - \boldsymbol{R}^{3} \quad (5) \qquad ,$$

$$\boldsymbol{Z}_{Li} = \boldsymbol{R}^{1,2} \ / \left| \boldsymbol{R}^{1,2} \right| \quad (6) \qquad , \qquad \mathbf{X}_{Li} = (\boldsymbol{R}^{1,3} \times \boldsymbol{R}^{1,2}) \ / \left| \boldsymbol{R}^{1,3} \times \boldsymbol{R}^{1,2} \right| \quad (7) \quad \text{and} \quad \mathbf{Y}_{Li} = \boldsymbol{Z}_{Li} \times \mathbf{X}_{Li} \quad (8).$$

There are a few advantages to the rigid-segment approach. First, the rigid segment may include as many markers (more than three) as necessary. Obviously the more markers involved in the calculation of the Local Reference System the less noise. Second, the arrangement of the markers can be arbitrary, as long as they are not all connected in a line. This advantage gives great freedom and flexibility to suit an individual's research need. Third, since the markers are rigidly connected, concern about the relative intermarker movement is eliminated. As a result, a relatively reliable and accurate Local Reference System is available. However, the reference system that is established by this approach may lack anatomical representations.

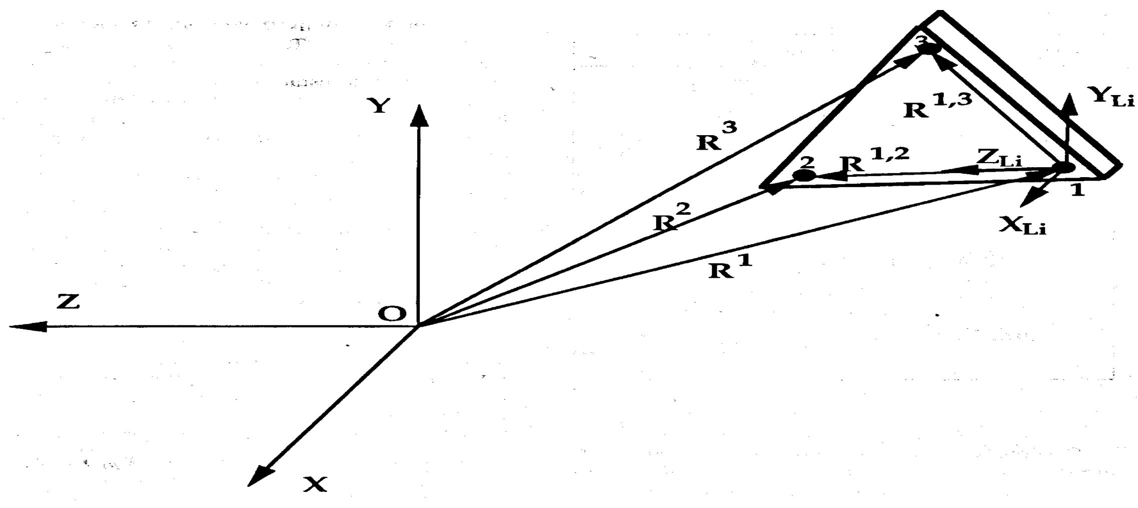


Figure 3.4 A rigid segment with three markers (Adapted from Craik R. and Oatis C., 1995).

Segmental Reference System

The Segmental Reference System is also called Anatomical Reference System. This system uses Cartesian coordinates that are not only fixed to the body segment, but also have clear anatomical meanings such as proximal-distal, medial-lateral and anterior-posterior.

Definition: A right-handed orthogonal triad $\langle X_{Si}, Y_{Si}, Z_{Si} \rangle$ fixed to a point on the ith body segment. The directions of the axes are defined as:

+ X_{Si} axis pointing anteriorly + Y_{Si} axis pointing proximally

+ Z_{Si} axis pointing laterally for the right side of the body segments and medially for the left side of the body segments

Needless to say, in order to have the axes represent the anatomical directions, markers that represent anatomical landmarks should be used to define the Segmental Reference System. For example, the landmarks on the shank may include medial and lateral malleolus, tibial tubercle, and femoral epicondyle, and the ones on the foot may include metatarsal heads (first through fifth), heel, and malleolus. Table 3.1 lists some widely used landmarks for pelvis, thigh, shank, and foot.

Obviously, the definition of the Segmental Reference System depends heavily on how to choose the landmarks and on how to construct the three orthogonal unit vectors based on the chosen landmarks. Because there is no standard yet on these issues, a consistent reference system for every body segment is not available. Nevertheless, a rule of thumb is that no matter how the landmarks are chosen, three points are the minimum number for each body segment in a three-dimensional case, and the three unit vectors should be determined in such a way that they are perpendicular to each other.

Now, let us consider a free body segment is (e.g., a shank) moving in space with an Absolute Reference System $\langle X, Y, Z \rangle$. Three landmark points a, b, and c are chosen from the segment: a is at medial malleolus, b is at lateral malleolus, and c is at tibial tubercle. Their locations in the Absolute Reference System are known as $\boldsymbol{R}^{a}$, $\boldsymbol{R}^{b}$, and $\boldsymbol{R}^{c}$ (see Figure 3.5). Based on these three position vectors we are ready to establish a segmental Reference System $\langle \mathbf{X}_{Si}, \mathbf{Y}_{Si}, \mathbf{Z}_{Si} \rangle$ in the following way:

$$\boldsymbol{R}^{a,b} = \boldsymbol{R}^{a} - \boldsymbol{R}^{b} \quad (9) \quad , \qquad \boldsymbol{R}^{a,c} = \boldsymbol{R}^{a} - \boldsymbol{R}^{c} \quad (10)$$

$$\boldsymbol{Z}_{Si} = \boldsymbol{R}^{a,b} \,/\, \left| \boldsymbol{R}^{a,b} \right| \quad (11) \quad , \qquad \mathbf{X}_{Si} = (\boldsymbol{R}^{a,c} \times \boldsymbol{R}^{a,b}) \,/\, \left| \boldsymbol{R}^{a,c} \times \boldsymbol{R}^{a,b} \right| \quad (12) \quad \text{and} \quad \mathbf{Y}_{Si} = \boldsymbol{Z}_{Si} \times \mathbf{X}_{Si} \quad (13).$$

Based on Equations 9 through 13 and the landmarks as listed in Table 3.1, a set of Segmental Reference Systems for the pelvis, thigh, shank and foot can be established.

Overall, the main advantages of the anatomical landmark approach are that some of the axes may have anatomical meanings (i.e., along the anatomical directions of the bones) and that the localization of landmarks is less dependent on personal judgment.

Body segment	Landmarks
Foot	1. Metatarsal heads
	2. Heel
	3. Medial malleolus
	4. Lateral malleolus
Shank	1. Medial malleolus
	2. Lateral malleolus
	3. Tibia tubercle
	4. Lateral femoral epicondyle
Thigh	1. Medial femoral epicondyle
	2. Lateral femoral epicondyle
	3. Greater trochanter
Pelvis	1. Left anterior superior iliac spine (ASIS)
	2. Right ASIS
	1. Left posterior superior iliac spine (PSIS)
	2. Right PSIS
	3. Sacrum

Table 3.1 Commonly used landmarks from the body for defining segmental reference systems.

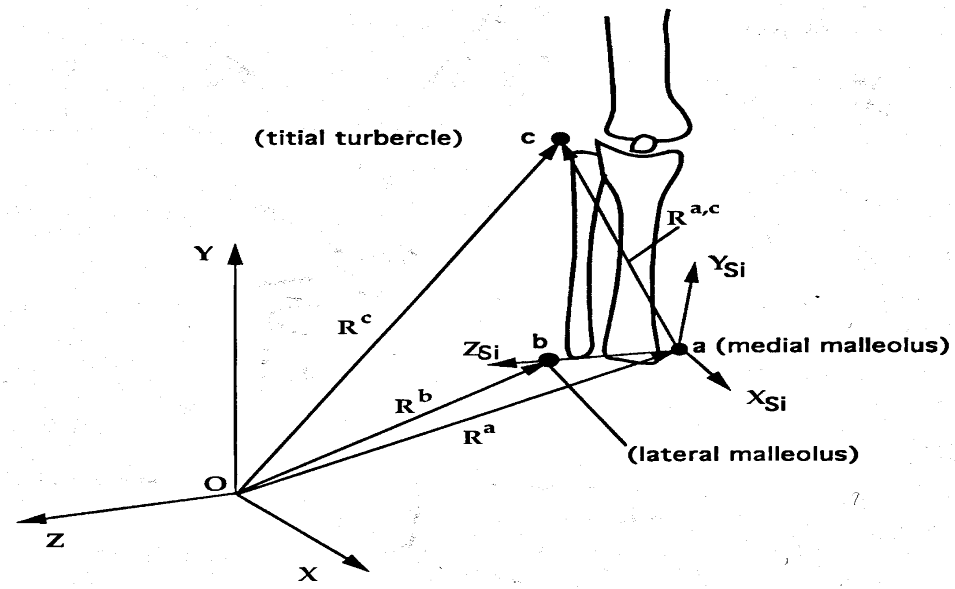


Figure 3.5 Conceptual diagram of determining the Segmental Reference System via landmark approach (Adapted from Craik R. and Oatis C., 1995).

Joint Reference System

A Joint Reference System is a system that is fixed to a joint. It is needed in order to describe the relative movement of the body segments with respect to each other. It allows rotations about axes with anatomical meanings such as flexion-extension, inversion-eversion and adduction-abduction, which are in common usage in clinical medicine. However, such a system may consist of nonorthogonal axes, but as long as force and moment are not resolved along these nonorthogonal axes, this does not present a problem.

Definition: A right-handed triad $<\mathbf{X}_{Ji}, \mathbf{Y}_{Ji}, \mathbf{Z}_{Ji}>$ fixed to a point in the i^{th} joint that connects the i^{th} and the $(i + 1)^{th}$ segments. The axes are defined as:

$\mathbf{X}_{Ji}$ axis representing an axis of the $(1 + l)^{th}$ Segmental Reference System

$\mathbf{Z}_{Ji}$ axis representing an axis of the i^{th} Segmental Reference System

$\mathbf{Y}_{Ji}$ axis representing a floating axis that is the cross product of $\mathbf{Z}_{Ji}$ and $\mathbf{X}_{Ji}$.

The Joint Reference System is defined for each joint individually. The most well-known examples of such systems are those developed for the knee by Grood and Suntay and Chao (see Figure 3.6). Two axes in the Segmental Reference Systems are established relative to anatomical landmarks, one in each body segment on opposing sides of the joint. The third axis, or the floating axis, that is perpendicular to each of the two segmental axes is then determined.

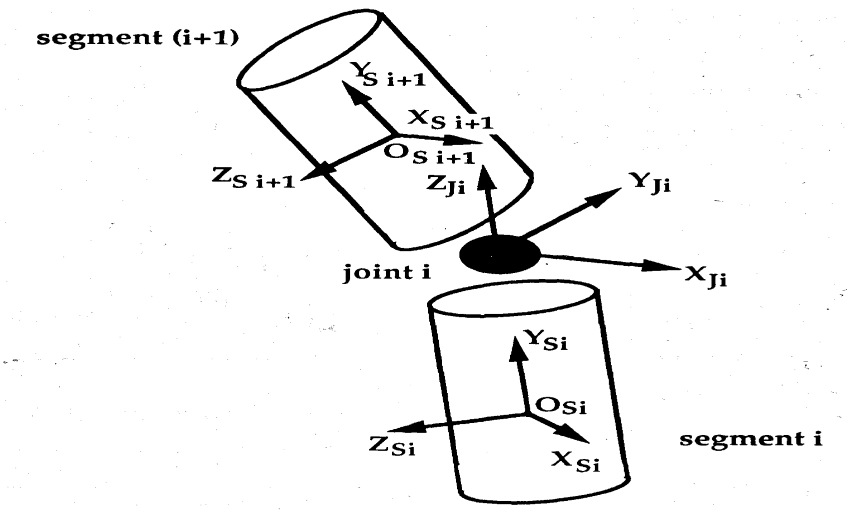


Figure 3.6 Joint reference system of the knee (Adapted from Craik R. and Oatis C., 1995).

3.3TRANSFORMATION BETWEEN REFERENCE SYSTEMS

Reference systems established for the purpose of describing the movement of a body segment. In general, the same movement may look quite different when it is observed in different reference systems. However, these different descriptions of a single movement are related by the transformation between the reference systems.

Linear Transformation

Without loosing generality, let us assume that a body segment ***i***, with a Local Reference System $<\mathbf{X}_{Li}, \mathbf{Y}_{Li}, \mathbf{Z}_{Li}>$, is moving in a space with Absolute Reference System $<\boldsymbol{X}, \boldsymbol{Y}, \boldsymbol{Z}>$ (see Fig. 3.7A). Originally, both reference systems coincide with each other. That is, the origins of the Absolute system ($\boldsymbol{O}$) and the Local system ($\boldsymbol{O}_{Lj}$), as well as the corresponding axes in both systems overlap. Given an arbitrary point ***p*** on the segment, its position in the Absolute Reference System $\boldsymbol{R}^{\boldsymbol{P}}$ and the Local Reference System $\boldsymbol{R}^{\boldsymbol{P}}_{\boldsymbol{Li}}$ can be described as:

$$\boldsymbol{R}^{\boldsymbol{P}} = [x^P\, y^P\, z^P]^{\mathrm{T}} \quad (14) \qquad \text{and} \qquad \boldsymbol{R}^{\boldsymbol{P}}_{\boldsymbol{Li}} = [x^P_{Li}\, y^P_{Li}\, z^P_{Li}]^{\mathrm{T}} \quad (15)$$

Clearly, at time $t = t_O$, we have: $\boldsymbol{R}^{\boldsymbol{P}} = \boldsymbol{R}^{\boldsymbol{P}}_{\boldsymbol{Li}}$ (16)

or

$$\begin{pmatrix} x^P \\ y^P \\ z^P \end{pmatrix} = \begin{pmatrix} x^P{}_{Li} \\ y^P{}_{Li} \\ z^P{}_{Li} \end{pmatrix} \quad (17)$$

Now, let us further assume that the segment ***i*** is only moving translationally, that is there is no rotational movement. At time $t = t_n$ *the* segment, along with the Local Reference System, moves to another location which is represented by the position vector of O_{Li} ($\boldsymbol{R}^{OLi}$), (see Figure 3.7B):

$$\boldsymbol{R}^{\boldsymbol{OLi}} = [x^{OLi}\ y^{OLi}\ z^{OLi}]^{\mathrm{T}} \quad (18)$$

Based on vector algebra, the position of point *p* that is described in the Local Reference System is related to the one in the Absolute Reference System in the following way:

$$\boldsymbol{R}^{\boldsymbol{P}} = \boldsymbol{R}^{\boldsymbol{P}}{}_{\boldsymbol{Li}} + \boldsymbol{R}^{\boldsymbol{OLi}} \quad (19)$$

or

$$\begin{pmatrix} x^P \\ y^P \\ z^P \end{pmatrix} = \begin{pmatrix} x^P{}_{Li} \\ y^P{}_{Li} \\ z^P{}_{Li} \end{pmatrix} + \begin{pmatrix} x^{OLi} \\ y^{OLi} \\ z^{OLi} \end{pmatrix} \quad (20)$$

In order to generalize the description of the linear transformation, it is now necessary to introduce the quatrain or quaternion quantities method. A quatrain is a four-element vector including a spatial vector (three elements) and a scalar (one element). In general, the scalar is taken to be1. For example, the quatrain form of vector $\boldsymbol{R}^{\boldsymbol{P}}$ is:

$$\boldsymbol{R}^{\boldsymbol{P}} = [x^P\ y^P\ z^P\ 1]^{\mathrm{T}} \ (21)$$

Now rearranging equation 21 in quaternion format, we obtain:

$$\begin{pmatrix} \mathrm{x}^P \\ y^P \\ z^P \\ 1 \end{pmatrix} = \begin{pmatrix} 1 & 0 & 0 & x^{OLi} \\ 0 & 1 & 0 & y^{OLi} \\ 0 & 0 & 1 & z^{OLi} \\ 0 & 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} \mathrm{x}^P{}_{Li} \\ y^P{}_{Li} \\ z^P{}_{Li} \\ 1 \end{pmatrix} \quad (22)$$

We define a 4/4 matrix $Ş_{Li}$ as:

$$Ş_{Li} = \begin{pmatrix} 1 & 0 & 0 & x^{OLi} \\ 0 & 1 & 0 & y^{OLi} \\ 0 & 0 & 1 & z^{OLi} \\ 0 & 0 & 0 & 1 \end{pmatrix} \quad (23)$$

which is called the linear transformation matrix from the Local Reference System *i* to the Absolute Reference System.

Expanding the preceding example into a general case, the linear transformation from Reference System II to Reference System I is:

$$\begin{pmatrix} x^P_I \\ y^P_I \\ z^P_I \\ 1 \end{pmatrix} = \begin{pmatrix} 1 & 0 & 0 & x_{\mathrm{I}}^{O\,\mathrm{II}} \\ 0 & 1 & 0 & y_{\mathrm{I}}^{O\,\mathrm{II}} \\ 0 & 0 & 1 & z_{\mathrm{I}}^{O\,\mathrm{II}} \\ 0 & 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} x^P_{\mathrm{II}} \\ y^P_{\mathrm{II}} \\ z^P_{\mathrm{II}} \\ 1 \end{pmatrix} = Ş_{\mathrm{I,\,II}} \begin{pmatrix} x^P_{\mathrm{II}} \\ y^P_{\mathrm{II}} \\ z^P_{\mathrm{II}} \\ 1 \end{pmatrix} \quad (24)$$

where $[x^k_j \;\; y^k_j \;\; z^k_j \;\; 1]^T$ represents the coordinates of point *k* (*k* is either an arbitrary point *p* or the origin of the *J* Reference System O_J) in the *J* Reference System (J = I or II).

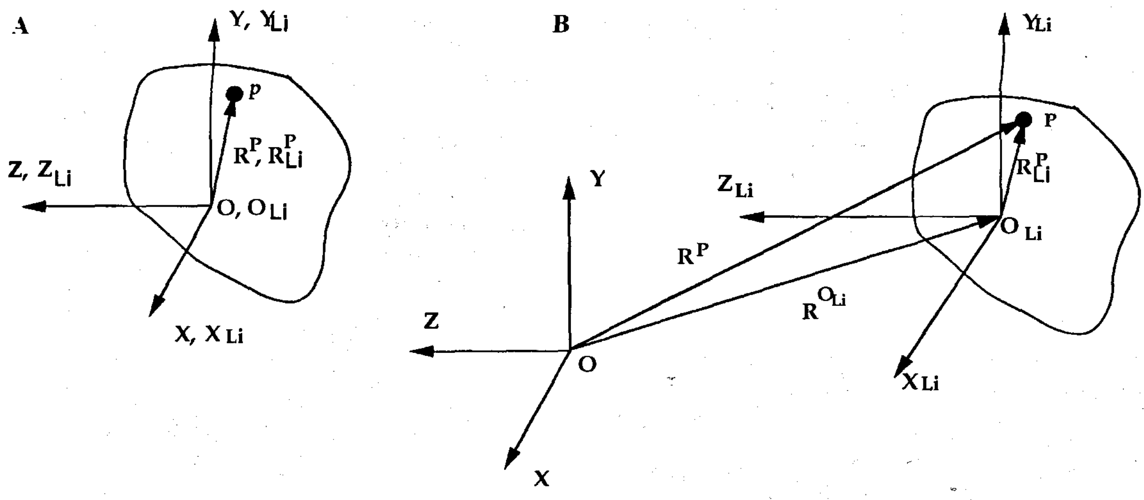


Figure 3.7 Linear transformation: A at $t = t_0$, both Absolute and Local Reference Systems coincide each other and B at $t=t_n$, Local Reference System translates to another location (Adapted from Craik R. & Oatis C., 1995).

3.4DEFINITION OF KINEMATIC VARIABLES

A free rigid body moving in a three-dimension space has a total of six degrees of freedom: three linear movements and three rotational movements. The status of such movement can be completely described by the following variables: displacement, velocity, and acceleration (Winter, 2009).

Displacement

Displacement describes the change in position of the body relative to a reference system. The change that is in a translational fashion is called linear displacement, whereas the change that is in a rotational fashion is called angular displacement

When the displacement of a body segment is presented relative to the Absolute Reference System, it describes the absolute movement of the body. However, such a description rarely has anatomical meanings. Usually, in gait analysis, the displacement of a body segment is presented either with respect to another body segment to describe their relative movement, or in a Joint Reference System to describe the joint movement. Nevertheless, no matter which reference system is used to present the displacement, it should be kept in mind that these representations are interrelated and transformable.

Velocity

The Velocity of a movement describes the speed of the change in position. For example, when a person moves from point A to point B in space, it could take 1 minute to complete the move if the person moves quickly or 10 minutes if the person moves slowly. The velocity is defined as the amount of change in position per unit time. If the position change is linear, the velocity is called the linear velocity. Likewise, if the position change is in rotation, the velocity is called angular velocity.

It should be stressed that the linear velocity depends on the location on the body segment, whereas the angular velocity does not. For example, during gait, the velocities at the proximal and distal ends of the shank are, for most of the time, different. However, the linear velocities at two different locations are closely related. The relation is expressed by the following equation:

$$\boldsymbol{V^p_L} = \boldsymbol{V^q_L} + \boldsymbol{\Omega_L} \times \boldsymbol{R^{p,q}_L} \quad (25)$$

where $\boldsymbol{V^p_L}$ and $\boldsymbol{V^q_L}$ are the absolute linear velocity vectors at points *p* and *q* on one body segment, respectively: $\boldsymbol{\Omega_L}$ is the angular velocity vector of the body segment; and $\boldsymbol{R^{p,q}_L}$ is the position vector between these two points. The subscript *L* indicates the variable is expressed in the Local Reference System. Therefore, as long as the velocity at one point of the rigid body is known, the velocities at other locations of the body can be determined.

Acceleration

Acceleration describes the speed of change in velocity. It is defined as the change in velocity per unit time. As with displacement and velocity, there are linear and angular accelerations as well. Also, the linear acceleration is location dependent. The relation between the linear accelerations at two arbitrary points on a body segment is as follows:

$$\boldsymbol{a^p_L} = \boldsymbol{a^q_L} + \boldsymbol{a_L} \times \boldsymbol{R^{p,q}_L} + 2\,\boldsymbol{\Omega_L} \times \frac{\boldsymbol{d\,R^{p,q}_L}}{\boldsymbol{dt}} + \boldsymbol{\Omega_L} \times (\boldsymbol{\Omega_L} \times \boldsymbol{R^{p,q}_L}) \qquad (26)$$

where $\boldsymbol{a^p_L}$ and $\boldsymbol{a^q_L}$ are the linear accelerations at points *p* and *q*, respectively, and $\boldsymbol{a_L}$ is the angular acceleration of the body segment.

3.5PHOTOGRAMMETRY: A BASIS FOR IMAGING ANALYSIS

General Principles and errors

Photogrammetry consists of obtaining measurements of three-dimensional (3-D) objects based on two-dimensional (2-D) images of the object while stereophotogrammetry consists of two or more 2-D spatially unique images, or targets, producing 3-D object. Typical close-range stereophotometric systems consist of two or more imaging devices positioned at fixed locations in a globa1ly referenced coordinate system (GRCS) with the imaging devices' axes forming mutually oblique angles (see Figure 3.8). Determination of the 3-D location, or 3-D coordinates, of an object from multiple 2-D images can be broken into three specific tasks: (1) identification of the object or target, (2) determination of the center or centroid, of the target, and (3) triangulation using the multiple 2-D images to identify the 3-D coordinates of the target. The 3-D reconstruction of the target's location utilizes the azimuth and elevation angles from each of the two or more cameras to extract the three spatial coordinates. The most popular 3-D reconstruction method is the Direct Linear Transformation (DLT), which uses the azimuth and elevation angles and the 2-D coordinates from each 2-D image (that is, from each camera) to derive the 3-D coordinates of the target. The DLT determines the relative camera locations and orientations, but does not correct for image distortions. A simplified introduction to Abdel-Aziz and Karara's DLT (1971) is presented in the appendix to this chapter. In addition to the "standard" DLT which typically uses 11 parameters, several modified DLTs exist each trying to avoid the nonlinear or nonorthogonality problems.

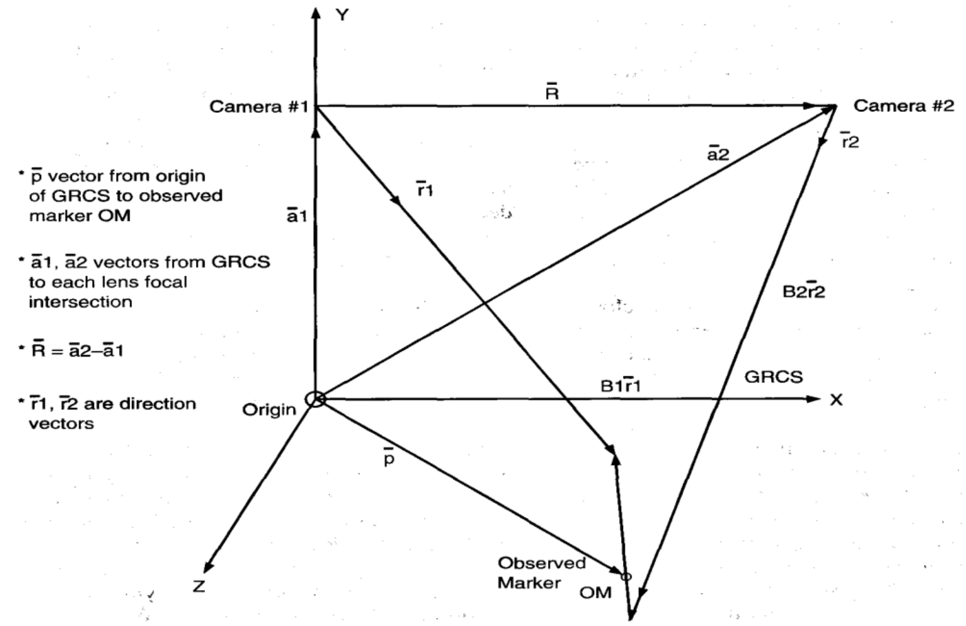


Figure 3.8 Globally referenced coordinate system (GRCS), (Adapted from Craik R. and Oatis C., 1995).

All measurement systems, including the kinematic systems to be described, suffer from measurement errors. Measurement accuracy depends to a large extent on the field of view of the cameras, although it also differs somewhat between the different systems. The earlier systems had measurement errors of 2-3mm in all three dimensions, throughout a volume large enough to cover a complete gait cycle (Whittle 1982). Recently, deign and (especially) calibration improvements have reduced typical errors to less than 1mm.

some commercial systems claim to provide much higher accuracy than this, but according to Whittle such claims are treated with skepticism, particularly when applied to the measurement a moving markers under realistic gait laboratory conditions.

Technical descriptions of kinematic systems use, and sometimes misuse, the terms ‘resolution’, ‘precision’ and ‘accuracy’. In practical terms, resolution means the ability of the system to measure small changes in marker position. Precision is a measure of system ‘noise’, being based on the amount of variability there is between one frame of data and the next.

For the majority of users, the most important parameter is *accuracy*, which describes the relationship between where the markers really are and where the system says they are!

Most commercial systems are sufficiently accurate to measure the positions of the limbs and the angles of the joints. However, the calculation of linear or angular velocity requires the mathematical differentiation of the position data, which magnifies any measurement errors. A second differentiation is required to determine acceleration, and a small amount of measurement `noise' in the original data leads to wildly erratic and often unusable results for acceleration. The usual way of avoiding this problem is to smooth the position data, using a low-pass filter, before differentiation. This achieves the desired object, but means that any genuinely high acceleration, such as that at the heel strike transient, are lost.

Thus kinematic systems are good at measuring position, but poor determining acceleration because of the problems of differentiating even slightly noisy data. Conversely, accelerometers are good at measuring acceleration, but poor at estimating position because of the problems of integrating data with baseline drift. Really accurate data could be obtained by combining the two methods, using each to correct the other and calculating the velocity from both. Some research studies have been conducted using this combined approach.

As well as the errors inherent in measuring the positions of the markers1 further errors are introduced because considerable movement may take place between a skin marker and the underlying bone. A few studies have been performed (e.g. Holden et al., 1997: Reinschmidt et al., 1997) in which steel pins have been inserted into the hones of volunteers (usual the investigators themselves) and the positions of skin markers compare with the positions of markers on the pins. The amount of skin movement revealed by such studies is generally somewhat worrying! The amount error this causes in the final result depends on which parameter is being measured. For example, marker movement has little effect on the sagittal plane knee angle because it causes only a small relative change in the length of fairly long segments, but it may cause considerable errors in transverse plane measurements or in measurements involving shorter segments, such as in the foot. In some cases, the magnitude of the error is greater than the measurement itself! Skin movement may also introduce significant errors in the calculation of joint moments and powers. A possibility for the future is to correct for marker movement by modeling the movement relative to the underlying bone.

A further error is introduced when the positions of the joints are estimated from anthropometric measures (e.g. leg length) and positions of skin markers, particularly where it is possible to place markers in the wrong position. Even for subjects with normal anatomy these errors can be substantial; for patients with bony deformity the errors may be even greater. This is a field in which improvements continue to be made.

There are two fundamentally different approaches to positioning markers on the limbs. One method is to mount each marker directly on the skin, generally over a bony anatomical landmark. The position and orientation of the limb segment is then defined by the marker positions. The other approach is to fix a set of al least three markers to each limb segment, either directly or mounted on a rigid structure (sometimes called a pod), so that its position and orientation can be determined in three-dimensional space. The movement of one limb segment relative to the next, and the position of the joint center, may then be derived mathematically. Both methods have advantages and disadvantages, and both suffer from errors due to marker movement. Rigid plates in particular may be moved by the contraction of the underlying muscles, and because they are relatively heavy their inertia causes them to lag behind the limb segment during rapid accelerations. Fig 3.9 shows two possible ways of arranging lower limb markers, one utilizing pods on the thigh and shank, and one based on markers mounted over anatomical landmarks.

Mention has already been made of the use of force platforms for the measurement of postural sway. Kinematic systems may also be used to make this type of measurement in the standing individual.

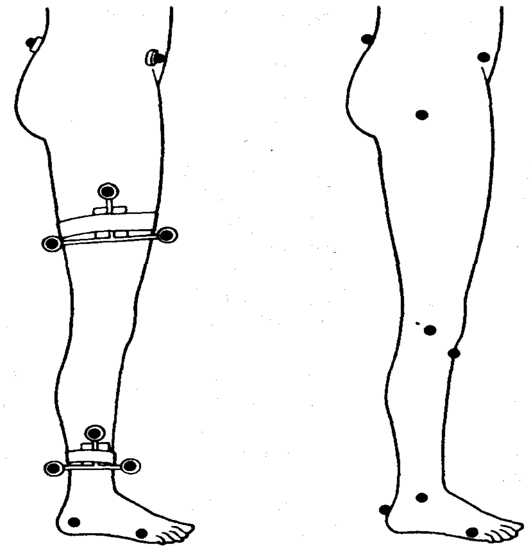


Figure 3.9 Typical marker configuration for the pelvis and the lower limb. Left: using rigid arrays (Cleveland Clinic). Right: skin-mounted, over anatomical landmarks (University of Oxford), (adapted from Whittle M. 2002).

Components

Imaging-based data acquisition systems are comprised of many intertwined hardware and software components. Each must be carefully integrated if the total system performance is to be optimized. A typical imaging system consists of the following components (Whittle M. 2002) :

1. Front-end optics consisting of a lens and detector

2. 2-D image acquisition, storage medium, and processing components consisting of either an online real-time analog-to-digital converter or an analog storage medium, such as a tape recorder, to an off-line digital converter, such as a frame grabber.

3. Image processing and geometric scaling component consisting of both hardware and software to convert the 2-D data into 3-D data. An abundance of off-the-shelf two- and three-dimensional systems exists, varying from inexpensive tape-recorded 2-D manual digitization to expensive, automated real-time 3-D systems. Imaging systems are either video-based or optoelectric. These systems are distinguished from each other by the kind of targets used. Video systems use passive targets, while optoelectric systems use active markers.

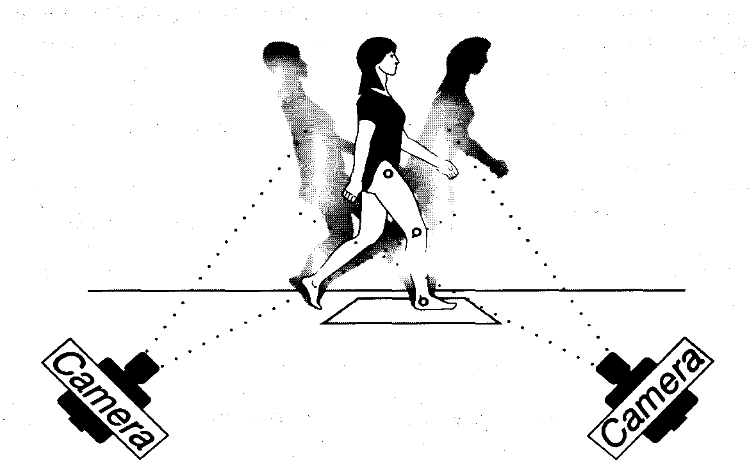


Figure 3.10 Two camera motion analyses provide a three-dimensional representation of the limb. Rotation does not alter the joint angles recorded (Whittle M. 2002).

3.6DIRECT MEASUREMENT OF KINEMATIC VARIABLES

Direct Measurement of Displacement

Perhaps the direct measurement of displacement of body segment (including linear and angular displacement) is the most popular approach in the field of gait analysis. There are many commercially available kinds of equipment, such as goniometry and video imaging techniques, for such purposes.

Direct Measurement of Acceleration

Direct measurement of acceleration requires the use of accelerometers. Although some accelerometers measure angular acceleration, most of them are for linear acceleration measurements.

Although the measurements of both linear displacement and linear acceleration are location dependent, the linear acceleration measurement is also orientation dependent under the influence of the field of gravity. For example, while an accelerometer is held stationary with its sensitive axis pointing vertically downward (i.e., along the direction of the gravity), its reading corresponds to 9.8 m/s^2 (or 1g). If it is rotated 90^0 so that its sensitive axis is perpendicular to the line of gravity, its reading is 0. Clearly, the output from the accelerometer depends heavily on its orientation.

Of course, such a problem may be avoided by using an accelerometer with a dynamic range above 0 *Hz* so that it is not sensitive to constant acceleration. However, this implies that not only the constant gravity acceleration will be eliminated, but some other constant accelerations that might be important in describing the movement of a body segment will be eliminated as well,

In addition to the orientation problem, the accelerometry approach is also concerned with the weight and size of accelerometers and with the vibration of soft tissue. Therefore, its application in the field of gait analysis is limited.

Direct Measurement of Multiple Variables

Each of the two previously mentioned direct measurements involves either displacement or acceleration. In order to derive other kinematic variables (such as velocity, etc.) based on these measurements, either differentiation or integration computation is needed (see the following section on Calculation of Kinematic Variables). Another approach, called the Integrated Kinematic measurement, has been used to directly measure a set of kinematic variables such as displacement, angular velocity, and linear acceleration.

The goal of the Integrated Kinematic measurement is to make necessary direct measurements of certain kinematic variables in order to achieve maximal accuracy in defining completely the status of movement of a rigid body segment. It was proposed that the following variables be measured: the displacements of at least three points on a rigid body segment, the angular velocity of the segment, and the linear acceleration of one point on the body. For detailed derivation of the complete set of kinematics, refer to the publication by Wu and Ladin (1993).

Measurement Approaches

Body segmental kinematics during locomotion includes the linear and angular displacements, velocities and accelerations. These variables have been studied in the past using one of the following roaches: (1) differentiation approach, which directly measures the displacement and calculates its time derivatives by numerical differentiation; (2) accelerometry approach, which directly means the acceleration and calculates the velocity displacement by numerical integration; and (3) integrated approach, which directly measures the displacement, velocity, and acceleration.

Those groups using either the differentiation or accelerometry approaches tended to report the directly measured kinematic data (i.e., the displacement or the acceleration), while those groups using the integrated approach covered a broader range of kinematics.

CHAPTER 4
DISCUSSION-HYPOTHESIS

4.1INTRODUCTION

The vertebra column and its function is directly related and influenced from several basic movement stereotypes involving hip extension, hip abduction, trunk flexion, neck flexion, shoulder abduction, push-ups (for the lower, upper and middle fixators of the scapula) and finally respiration (law of free floating equilibrium). These stereotypes are formed from different muscles or muscular groups. The proper action of each movement stereotype is dependable upon the activation within a specific sequence of the muscles composing this stereotype in order mobility as well stability of the axial system to be preserved constantly. These stereotypes are also responsible for the creation of different kind of forces applied upon the human structures or tissues and thus these forces must be well balanced too.

Whenever the normal sequence of muscle activation within a movement stereotype is interrupted or changed then this condition will exert a direct or indirect influence on the relevant adjacent or not adjacent structures of the axial system (through muscle chains or primary and secondary points), causing disturbances or/and misbalances in the posture and locomotion and eventually will lead to a functional or a structural disorder or both (Soderberg, 1996).

Disorders such is the obliquity of pelvis (Perry, 1992), the leg length difference (Panjabi and White, 1990), the abnormal segmental postural characteristics of the trunk (Zabjek et al, 2005) and the influence of an asymmetrical body weight distribution (Genthon et al, 2005) that exist in scoliosis are factors that true will affect the locomotion too (Chen et al, 1998) as well as any kind of a pathological effect involving the postural system, that will produce abnormalities to the segments of the body that are closely related with walkingt. As a result the movement will be affected in a minor or major degree. The basic movements of the trunk are affected too and eventually the creation of factors such as the tightness, the elongation or shortening of the soft tissues surrounding the bony structures which are directly (or indirectly) connected with the vertebrogenic disorder or pain may lead to an asymmetrical body image. Perry (1992), suggest that pain, sensory loss, muscle weakness and deformity are factors highly involved in the pathokinesiology of gait and in scoliosis, as any other disorder of the axial system, these factors are true to exist.

The disorder of our interest for studying is moderate idiopathic scoliosis. Abnormal spinal curvatures in the frontal or sagittal plane of the thoracic and/or lumbar spine have proven difficult to prevent or treat. Idiopathic scoliosis, taking into account curves above 10^0, is present in 2-3% of the general population (Lonestien, 1984, Renshow, 1988). It is most common during late childhood, in particular in girls (Marieb, 1998) but the prevalence of small curves, less than 20°, is about equal in males and females. From the total population of patients suffering from this condition, 3% will need some kind of conservative treatment (Lonstein, 1995). It has been estimated that approximately 65% of scoliosis cases are idiopathic (Agabegi et al 2008). Scoliosis is associated with increased pain in adults of all ages, compared with control populations (Mayo et al, 1994, Weinstein et al, 2003). Furthermore, children and adults with mild to moderate curvatures may have reduced exercise capacity (Schwab et al, 2003, Weinstein et al, 2003).

4.2HYPOTHESIS AND AIMS

The existence of any deviation upon the human structures or tissues through moderate idiopathic scoliosis will cause a misbalance and an influence upon the proper distribution of forces acting on and around a joint, ligament, bone or muscle. The result of such a misbalance, will be an alteration of all physical quantities exerted from different segments of the body not only in the upper trunk but in the lower trunk too and changes will occur in the upper extremities as well as in the lower extremities and their relative joints. The joints of the lower extremities, which are involved in the gait cycle, from such an influence their functions will be propably altered during the gait cycle; and movement restriction or loss is expected to be observed.

This study is relative to the three dimensional analysis of the gait cycle of young adults suffering from moderate idiopathic scoliosis (MIS), in the lumbar or thoracolumbar part of the vertebra column. It is based upon the presentation of cases suffering from scoliosis, patients that went under an elaborate thorough of kinesiologic, anthropometric and kinematic analysis so as someone to be capable to extract useful information of how the locomotion and the posture corresponds to this kind of disorder. Scoliosis patients exhibit significantly impaired quality of life (Schwab et al, 2003) and young adults with MIS consist a population group with increased occupational and sports activities (Weinstein et al, 2003) and gait cycle is of great importance. Gait analysis is used to identify and treat (Lewit , 2000, Zabjeket al,2005) individuals with conditions affecting their posture and in terms their ability to walk.

The Kinematic analysis is trying to seek how the gait cycle of the lower extremities correspond to moderate idiopathic scoliosis and what kind of alteration will be exerted upon the physical quantities (linear displacement, linear velocity and linear acceleration) that the major 3 joints (hip, knee and ankle) of the lower extremities produce during locomotion. Expectable asymmetries, for our interest, during locomotion may either concern the function (kinematic point of view) of the center of gravity or may concern the creation of abnormal locomotion from the lower limbs, or both.

The general hypothesis: the scoliotic shape of the trunk will result in dynamical characteristics of whole body locomotion movement.

The aim of this work is:

- ***The purpose of this work is:*** To detect the biomechanical reflexion of moderate idiopathic scoliosis upon the major joints of the lower extremities and the center of gravity during gait cycle of young adults as well as the correspondence of these anatomical points due to an abnormal movement created always in comparison with the gait cycle of healthy people.

CHAPTER 5
A PRACTICAL APPROACH OF KINEMATIC ANALYSIS

5.1MATERIALS AND METHODS

For the purpose of this study thirty-five young adults (with similar anthropometric characteristics) of both sexes were selected and divided in two groups: a) Group A consisted from 20 young adults with moderate idiopathic scoliosis and inclusion criteria for this group were age not to be more than eighteen years and less than fifty (as we wanted to study established deformities in people with no degenerative spine so we arbitrarily set as limit fifty years of age), selection of lumbar and thoraco-lumbar curves (as these will probably have greater impact on pelvic locomotion) and selection of scoliosis curves between 20^{o} to 40^{o} (smaller curves might not influence gait and bigger are not so commonly met) and b) Group B with 15 healthy people without any known spinal deformity or disease (table1) and no limb length discrepancy more than 0,5 cm (as this could influence gait cycle pattern).

All patients had a right thoraco-lumbar or left lumbar primary structural curve because these curves are closely related to pelvis and in terms with the gait cycle of human beings. The average Cobb's angle in group A was $29,4^{0}$ and plumb line declination was 1,2cm. Characteristic deformities were the functional leg length difference and the obliquity of pelvis that participated in an asynchronous gait cycle that both extremities in group A presented. Every subject signed on and participated freely in the study, approved by the local ethics board. All subjects were submitted to a clinical, radiological(Cobb, 1948) and gait assessment. The gait cycle was divided according the convexity created from the scoliosis. The term "Ipsilateral" is used for the convexity side joints and "Controlateral" for the concavity side joints in group A. In group B the distal joints of the lower extremities are presented with average values from both sides (left-right) due to minimal differences found between them.

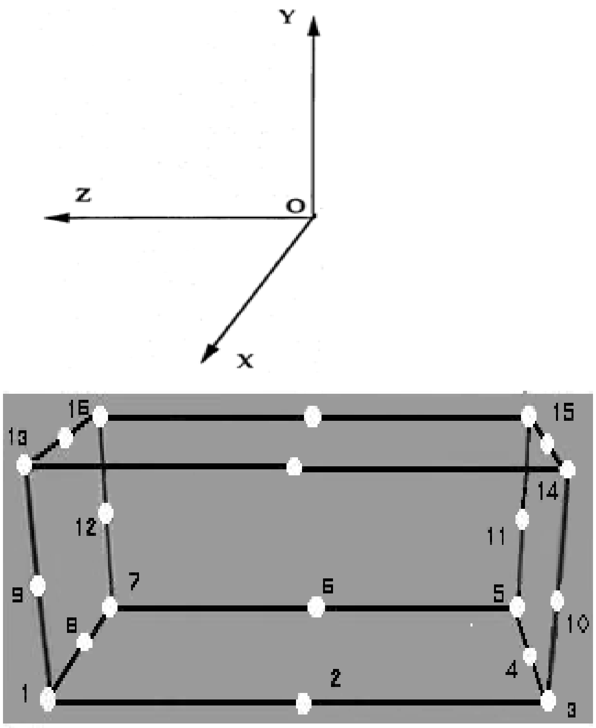
Demographic data	***Scoliosis patients*** Group A (*n*=20)	***Control group*** Group B (*n*=15)
Height (cm)	1,72 (1,55-1,90)	1,70 (1,57-1,91)
Body Weight (Kg)	74 (58-92)	72 (60-90)
Age (years)	32,4 (20-40)	36,1 (23-38)
Sex	12 Females 8 Males	8 Females 7 Males

Table 1 Demographic data

All the motion quantification systems depend on defining the arcs and positions of the individual joints numerically and thus on the lower extremities there was installation of paper markers on the skin surface of anatomical landmarks (Major trochanter, lateral condyle and lateral malleolus) that accurately represent the actions of specific joints (hip, knee and ankle) necessary for movement identification. The gait assessment was succeeded with three dimensional (3D) optical analysis of walking stereotypes with "forced walking" on a mechanical treadmill.

A mechanical treadmill was used and for 3D video motion analysis three digital video camera recorders where obtained. The record frequency was 50 half frames per second in each camera. From each video record were obtained coordinates of the above mentioned points and these coordinates were computed the kinematic parameters with respect to the weight and height of the patient during the locomotion on the treadmill.

A reference frame provided the reference for the description of the physical quantities of our interest (always with respect to the weight, height and the fixed anatomical points) which are: a) The linear displacement which defined as the position of a particle and its location at a given instance, b) linear velocity which defined as the rate of change in the position or the rate of displacement and c) linear acceleration which defined as the rate of change in velocity (Vaughan, Davis and O'Connor, 1999). Basically the reference frame displayed different perspectives in our view while the coordinate system provided different ways for the description of the physical quantities in these perspectives. This coordinate system was based upon the Cartesian 3D coordinate system (Figure 1).



	X axis	Y axis	Z axis
1	0.000	0.000	0.000
2	0.500	0.000	0.000
3	1.000	0.000	0.000
4	1.000	0.000	0.250
5	1.000	0.000	0.500
6	0.500	0.000	0.500
7	0.000	0.000	0.500
8	0.000	0.000	0.250
9	0.000	0.250	0.000
10	1.000	0.250	0.000
11	1.000	0.250	0.500
12	0.000	0.250	0.500
13	0.000	0.500	0.000
14	1.000	0.500	0.000
15	1.000	0.500	0.500
16	0.000	0.500	0.500

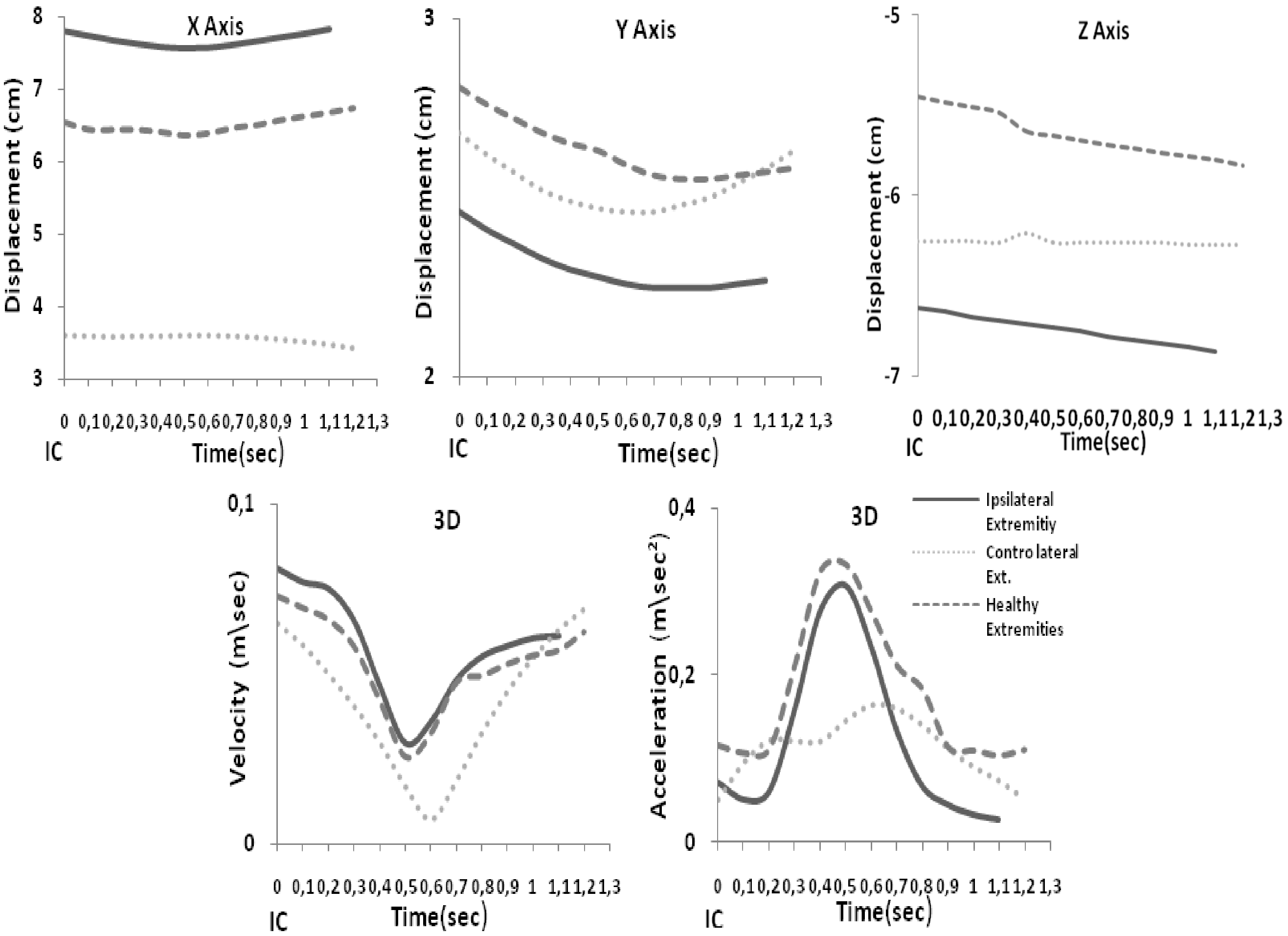
Fig. 1 A reference frame through a Cartesian 3D coordinate system and the coordinates of the control points.

Finally a light bulb was used to provide the necessary light as well a treadmill on which the patient was walking during the recording period and provided the steady base for the fixed cameras that were used.

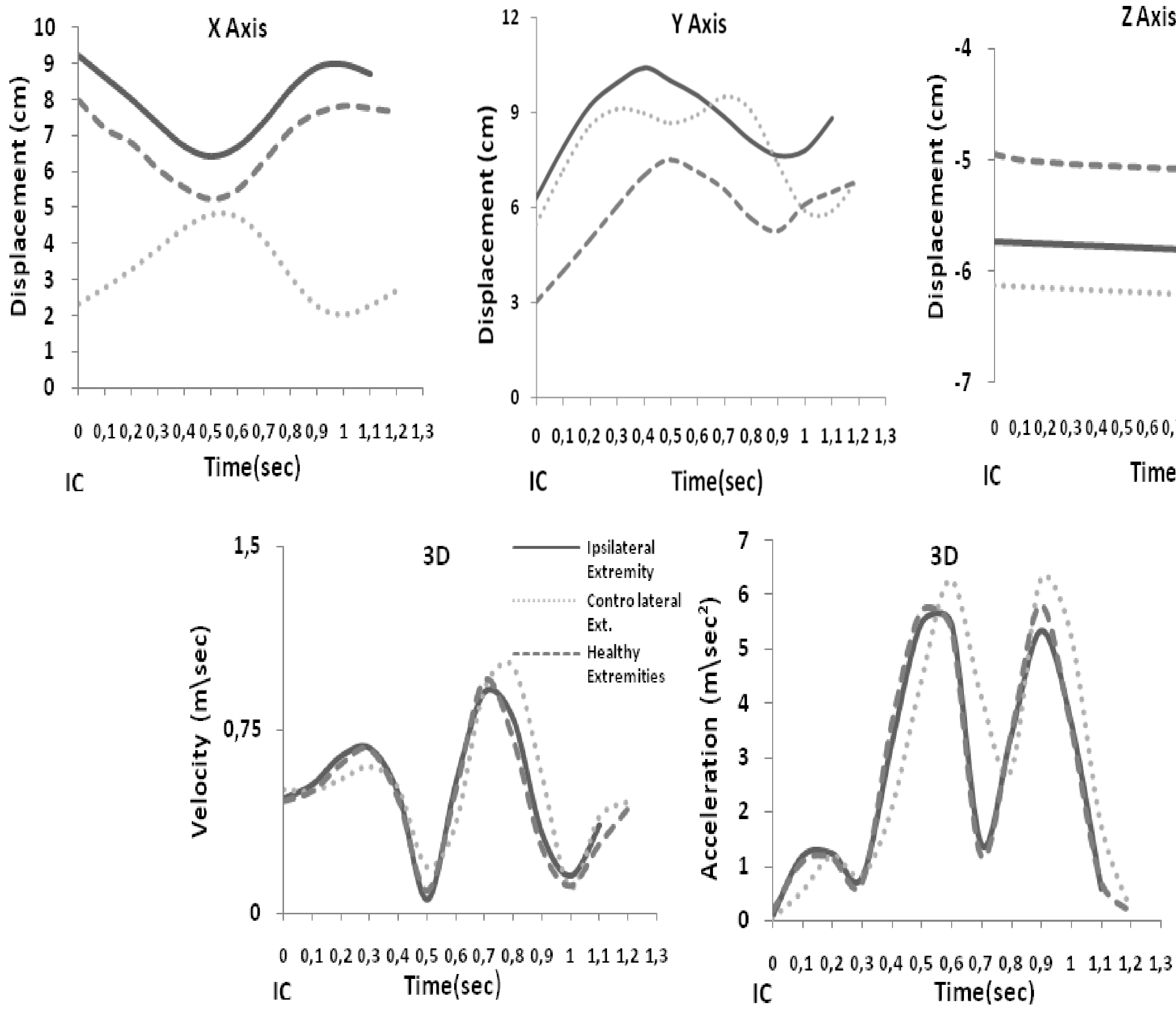
The 3D kinematic analysis of gait was succeeded with the subjects walked on a mechanical treadmill, with self selected speed and in terms closer to daily walking activities, while holding supporters from the device to prevent falling and performing several times the gait cycle. A typical one of the gait cycles was selected after achieving a steady pace (<1m/sec) of walking so as to avoid mistakes in measuring the physical quantities, and recorded by 3 fixed cameras that were placed in a circular manner and focused in the specific points of our interest and with respect to the coordinate system. Gait cycle in each extremity was initiated with initial contact (IC-heel contact) from double support position at 0 seconds and ended at the next initial contact so as to compare the movement stereotype between the lower extremities, and we recorded the locomotion of each leg separately. The recording took place from different angles so to be a 3D approach. Eventually the recorded data was being processed by the computer through software (Ariel Dynamics Software-APAS) and then a direct linear transformation (DLT Method) of the captured data was done so as to compute the three-dimensional image space coordinates of the subject's body joints from the relative two-dimensional digitized coordinates of each camera's view. This process involved the transformation of the relative digitized coordinates of each point in each frame to absolute image space coordinates through a computer and was possible to load all data into the computer while simultaneously displayed and evaluated multiple data sets for a desired sequence (Craik and Oatis, 1998). These measurements (Winter, 2009) allowed computation of the sagittal plane (x axis-forward / backward direction), vertical plane (y axis-gravitational-upward / downward direction), and frontal plane (z axis-left / right-medial / lateral direction) of the hip, knee and ankle joint, the center of gravity (CoG) as well as the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from these anatomical parts of the human body. Also the duration of the gait cycle (sec) and the angular degrees of freedom of the knee joint (sagittal plane) were computed.

For the purpose of statistical analyses power was set at 80%. Student paired t-test was used for the purpose of statistical analysis of the behavior of the knee and ankle joints across groups and sides of the body (side to side comparison) with level of significance at 95% (Confidence interval C.I 95%) and the mean value for each quantity was used for statistical analysis. The accepted significance level was <0.05 or lesser for all analyses.

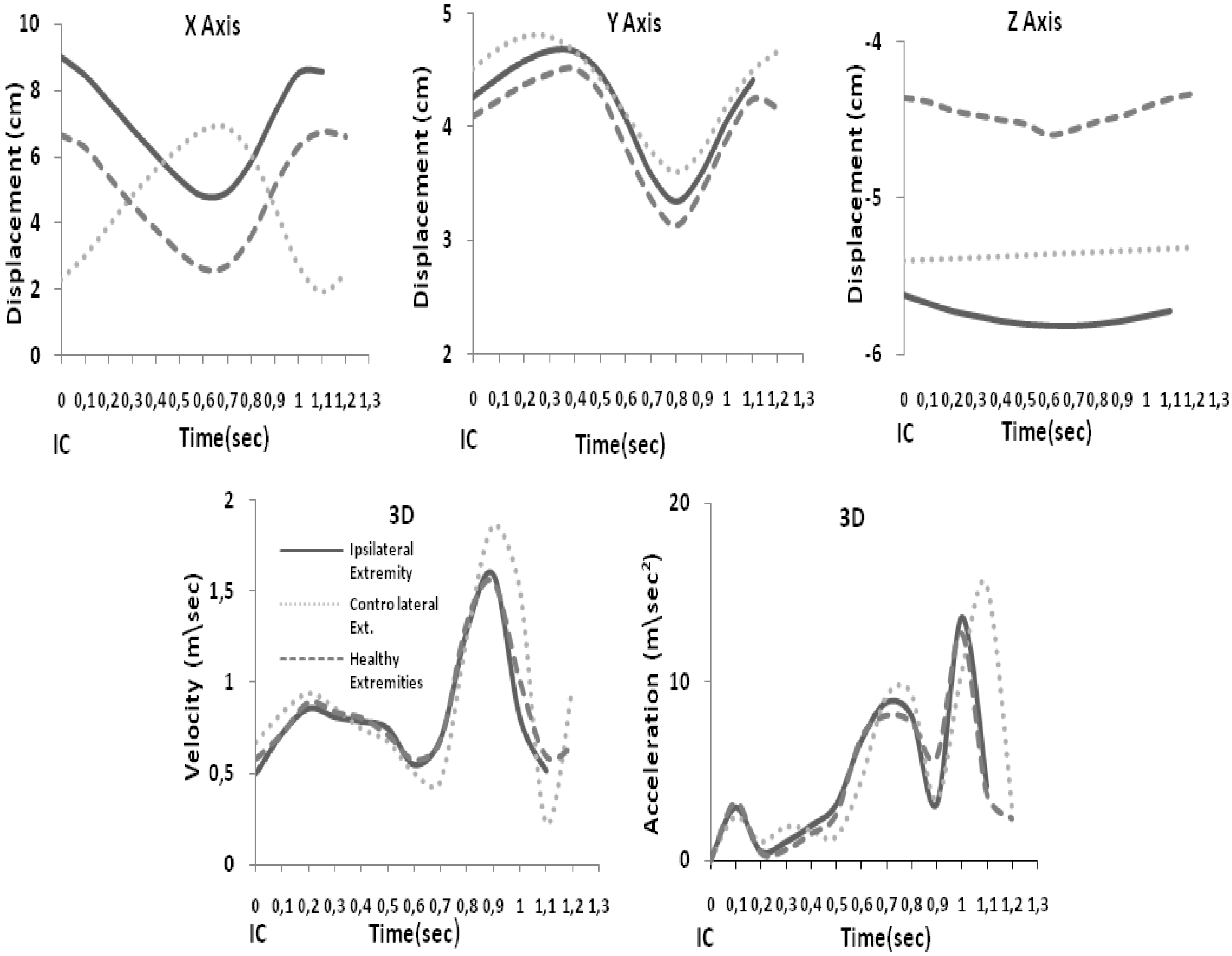
CHAPTER 6
KINESIOLOGIC AND KINEMATIC ANALYSIS-RESULTS



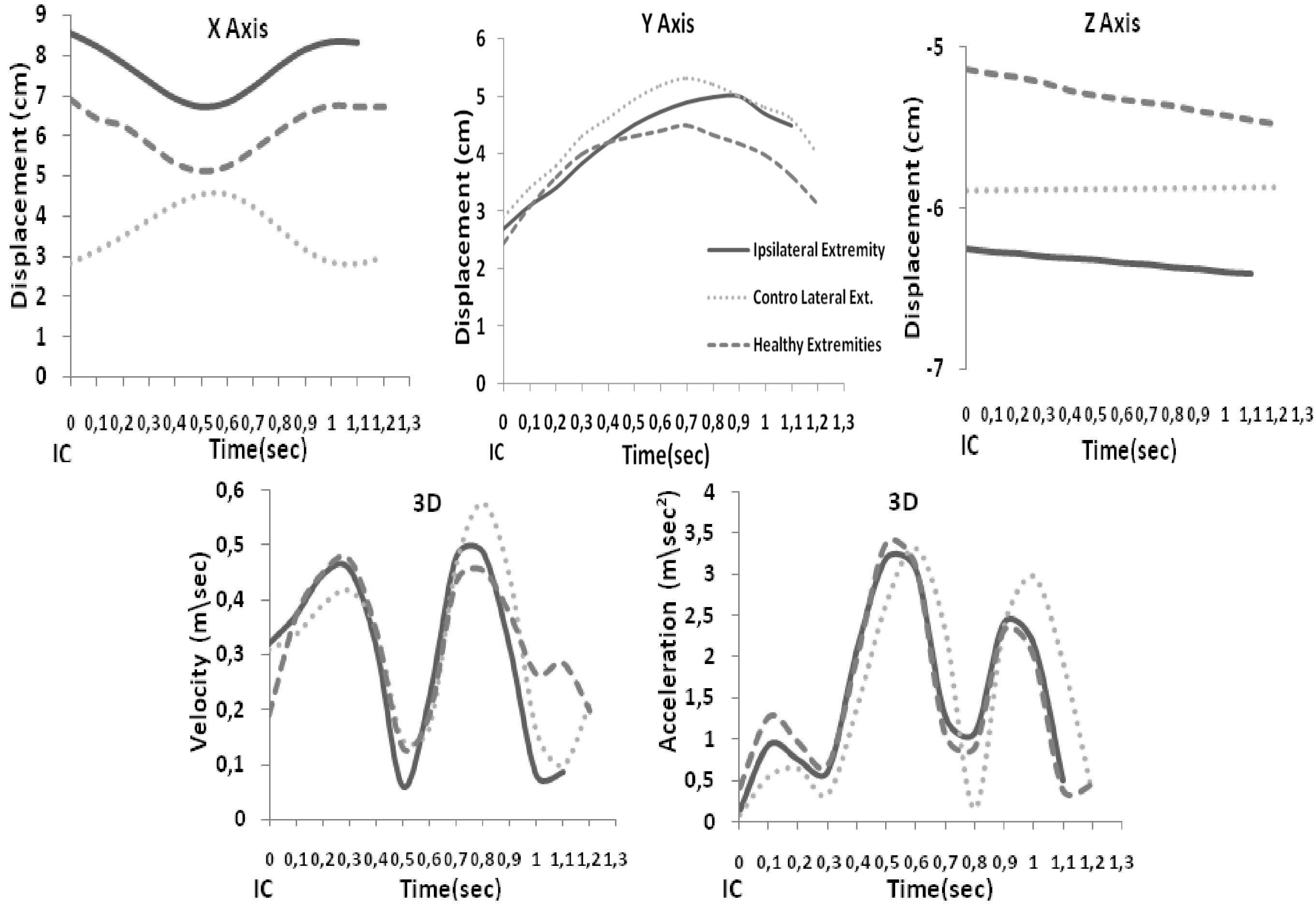
Graph 6.1 Thoracolumbar scoliosis: Typical linear displacement, velocity and acceleration of the hip joint during the gait cycle (IC stands for initial contact with the gait cycle initiated from double support phase) of young adults suffering from moderate idiopathic scoliosis (ipsilateral at the convex side) and healthy subjects (average from right and left extremity).



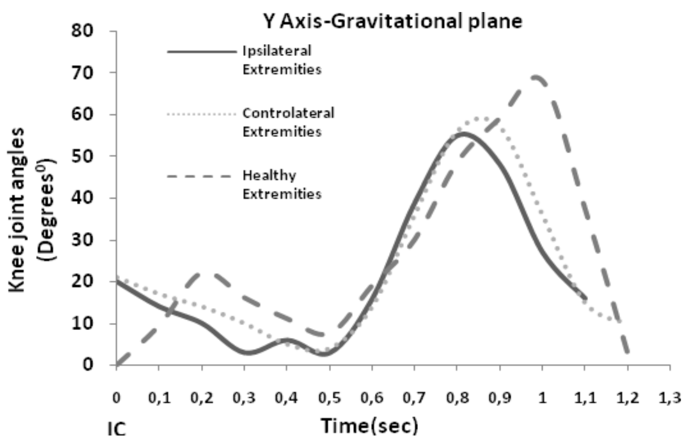
Graph 6.2 Thoracolumbar scoliosis: Typical linear displacement, velocity and acceleration of the knee joint during the gait cycle (IC stands for initial contact with the gait cycle initiated from double support phase) of young adults suffering from moderate idiopathic scoliosis (ipsilateral at the convex side) and healthy subjects (average from right and left extremity).



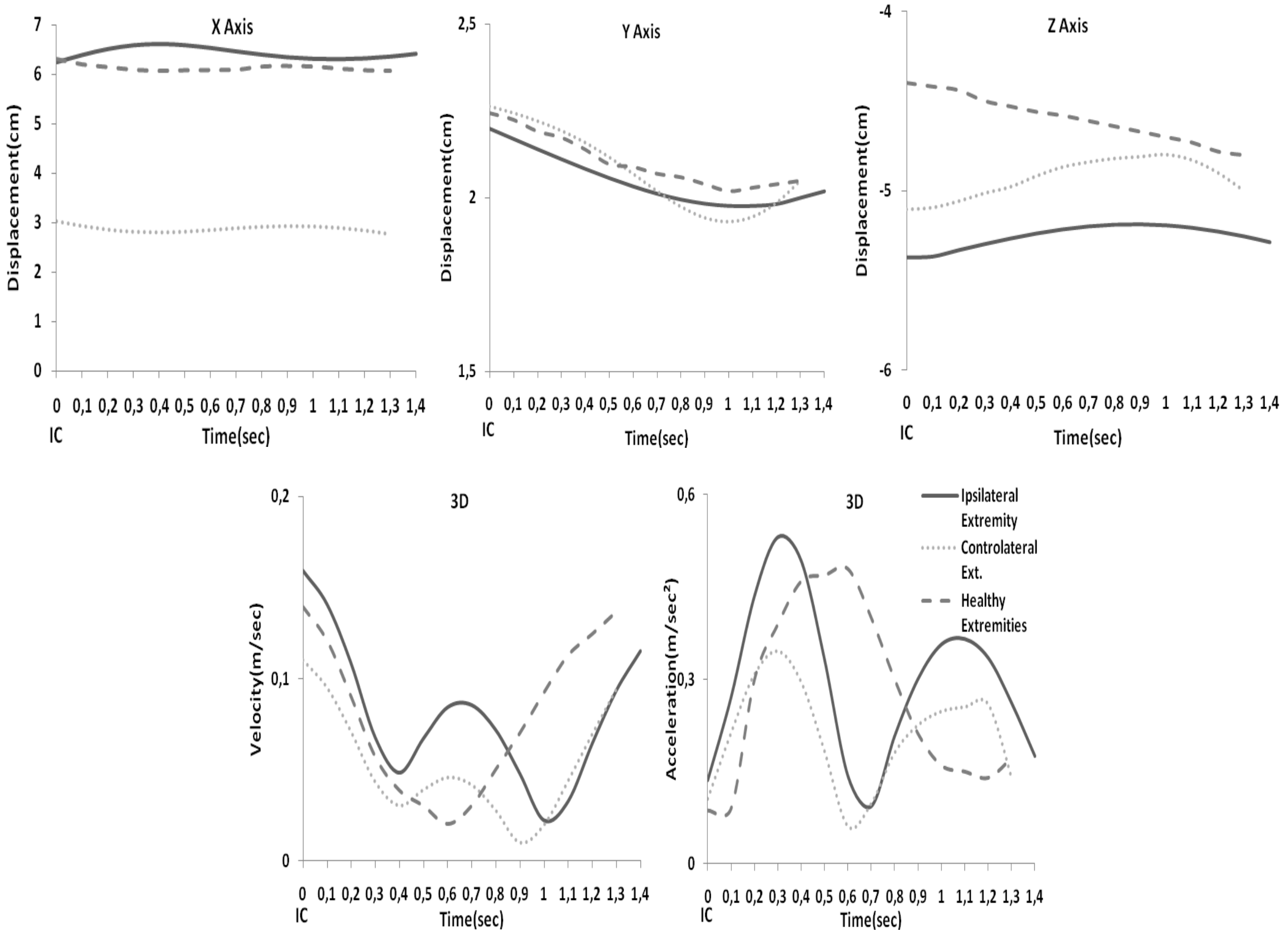
Graph 6.3 Thoracolumbar scoliosis: Typical linear displacement, velocity and acceleration of the ankle joint during the gait cycle (IC stands for initial contact with the gait cycle initiated from double support phase) of young adults suffering from moderate idiopathic scoliosis (ipsilateral at the convex side) and healthy subjects (average from right and left extremity).



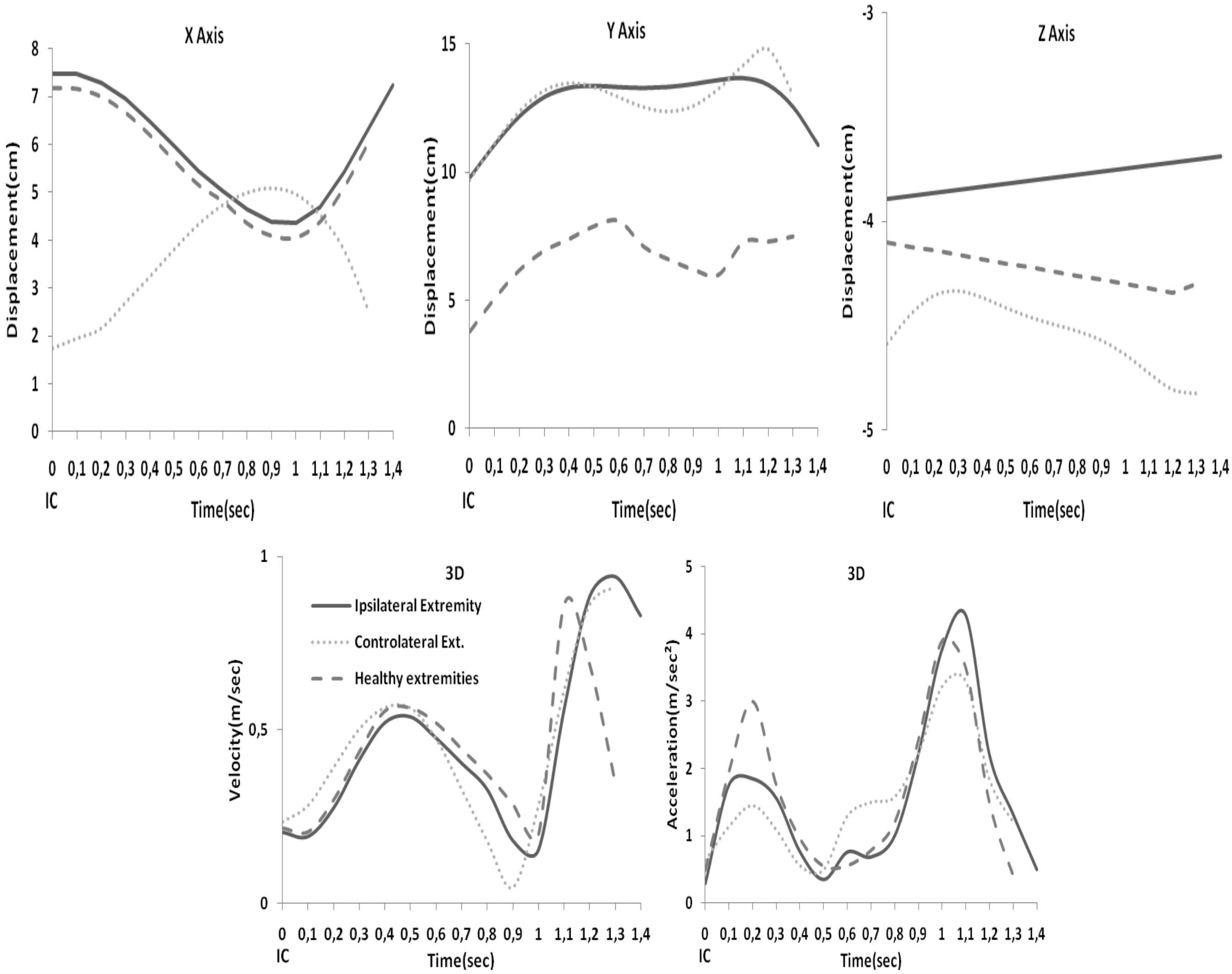
Graph 6.4 Thoracolumbar scoliosis: Typical linear displacement, velocity and acceleration of the center of gravity during the gait cycle (IC stands for initial contact with the gait cycle initiated from double support phase) of young adults suffering from moderate idiopathic scoliosis (ipsilateral at the convex side) and healthy subjects (average from right and left extremity).



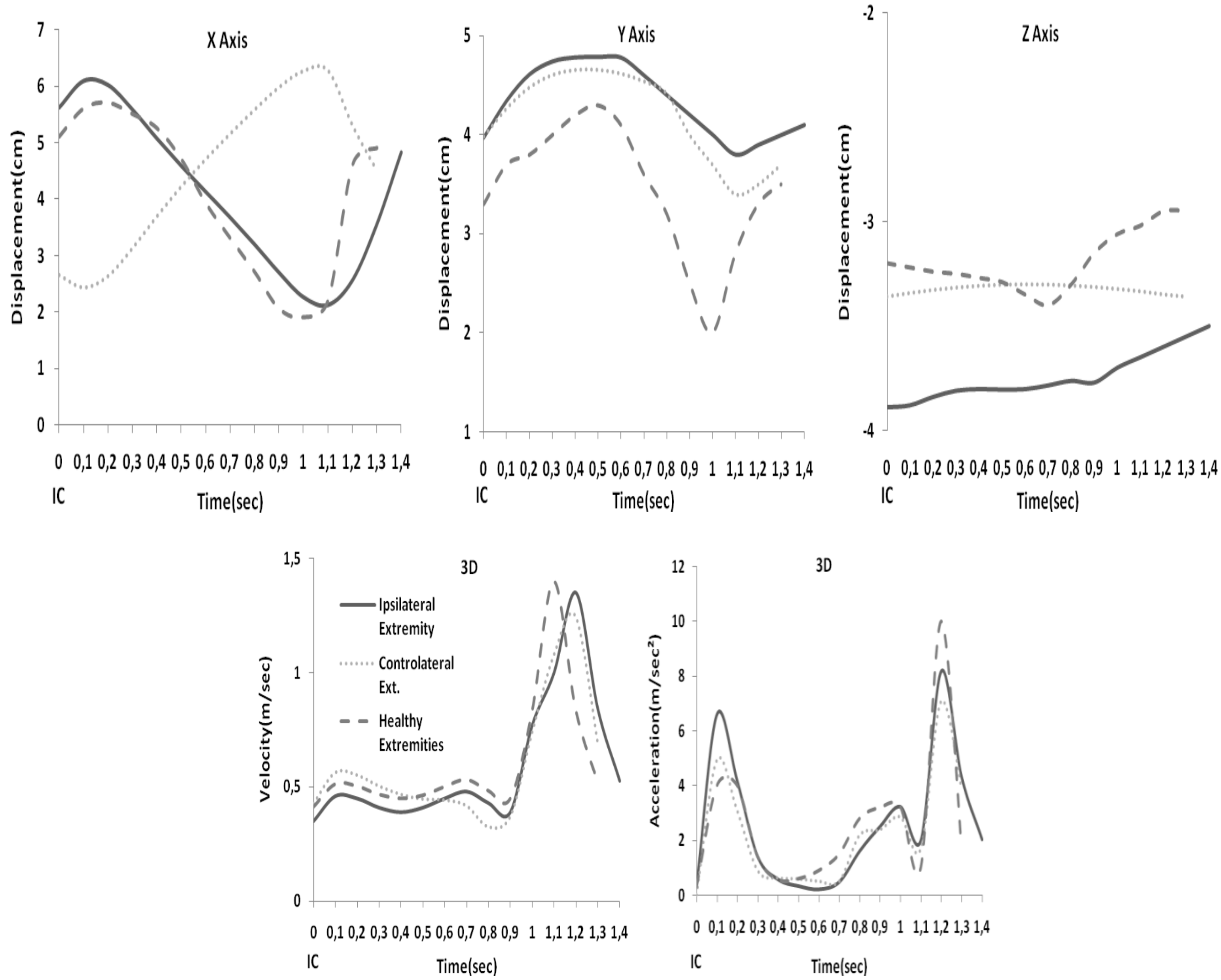
Graph 6.5 Thoracolumbar scoliosis: Typical angles of the knee joint during the gait cycle (IC stands for initial contact with the gait cycle initiated from double support phase) of young adults suffering from moderate idiopathic scoliosis (ipsilateral at the convex side) and healthy subjects (average from right and left extremity).



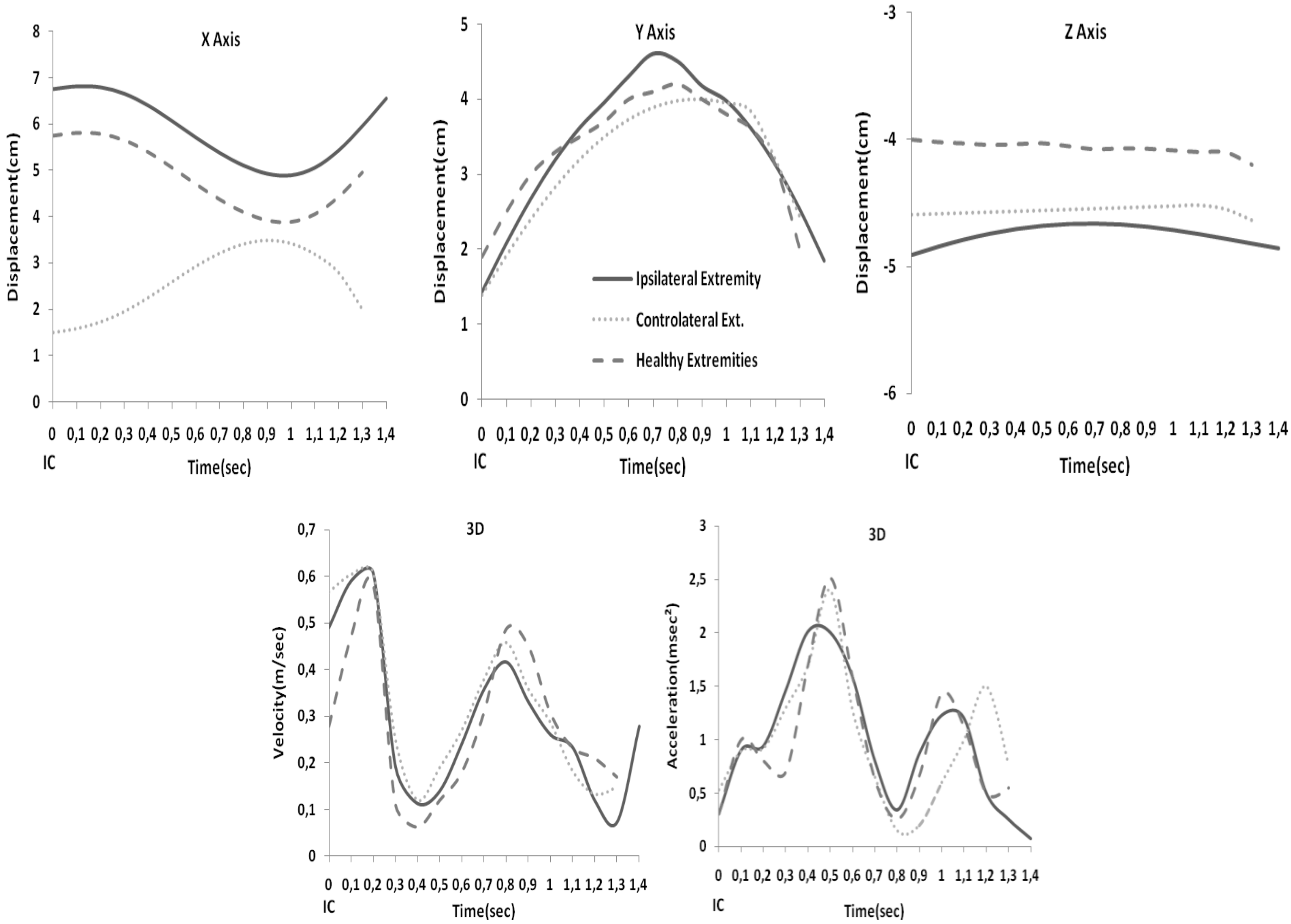
Graph 6.6 Lumbar scoliosis: Typical linear displacement, velocity and acceleration of the hip joint during the gait cycle (IC stands for initial contact with the gait cycle initiated from double support phase) of young adults suffering from moderate idiopathic scoliosis (ipsilateral at the convex side) and healthy subjects (average from right and left extremity).



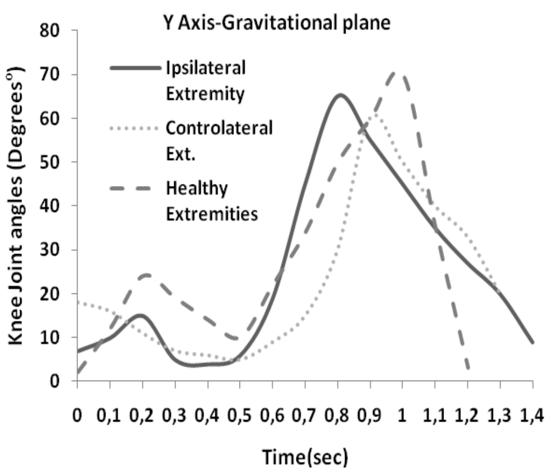
Graph 6.7 Lumbar scoliosis: Typical linear displacement, velocity and acceleration of the knee joint during the gait cycle (IC stands for initial contact with the gait cycle initiated from double support phase) of young adults suffering from moderate idiopathic scoliosis (ipsilateral at the convex side) and healthy subjects (average from right and left extremity).



Graph 6.8 Lumbar scoliosis: Typical linear displacement, velocity and acceleration of the ankle joint during the gait cycle (IC stands for initial contact with the gait cycle initiated from double support phase) of young adults suffering from moderate idiopathic scoliosis (ipsilateral at the convex side) and healthy subjects (average from right and left extremity).



Graph 6.9 Lumbar scoliosis: Typical linear displacement, velocity and acceleration of the center of gravity during the gait cycle (IC stands for initial contact with the gait cycle initiated from double support phase) of young adults suffering from moderate idiopathic scoliosis (ipsilateral at the convex side) and healthy subjects (average from right and left extremity).



Graph 6.10 Lumbar scoliosis: Typical angles of the knee joint during the gait cycle (IC stands for initial contact with the gait cycle initiated from double support phase) of young adults suffering from moderate idiopathic scoliosis (ipsilateral at the convex side) and healthy subjects (average from right and left extremity).

All patients of group A had a right thoraco-lumbar or left lumbar primary structural curve. The average Cobb's angle in group A was $29{,}4^0$and plumb line declination was 1,2cm. Mean leg length discrepancy in group A was 1,2cm (±0,2cm, C.I 95%) while in group B the difference was 0,3cm (±0,13).

Scoliosis patients showed typical deformities regarding lumbar region, scapula and pelvis (table 6.1). When examining range of motion, an obvious hip joint flexion restriction was noted as well as reduced lateral flexion of the spine ipsilateral to the curve. Restriction in rotation and extension of the spine was less, but worth noticing. More detailed clinical examination is shown in table 6.2.

Clinical data	Type of scoliosis	Lumbar Lordosis	Pelvic tilt	Head & neck posture	Shoulder & scapula position	Iliac crest & PSIS
Scoliosis patients Group A (*n*=20)	Thoraco-lumbar (16) Lumbar (4)	Hyper-lordosis in 60% of all cases. Flattening in 30 % of all cases.	Existed in 100% of all cases	Protruded in 100% of all cases	Elevation in 100% of all cases	Elevation in 100% of all cases
Control group Group B (*n*=15)	NE	Hyper-lordosis in 15% of all cases Flattening in 5 % of all cases	Existed in 5% of all cases	NE	NE	NE

Table 6.1 Effect of scoliosis upon the musculoskeletal system (NE: not existed).

Clinical data	Trunk and spine			Lumbar extension	Hip joint left & right extremity		SLR left & right extremity	Pain presence
	Flexion	Lateral flexion	Rotation	Range of movement	External rotation	Internal rotation	Hip flexion / knee extension	During movement
Scoliosis patients Group A (*n*=20)	Limited in 30% of all cases	Limited in one side in 100% of all cases	Limited in one side in 70% of all cases	Hypo-mobile in 80% of all cases	Difference in 100% of all cases	Difference in 100% of all cases	Difference in 85% of all cases	90% of all cases during lateral-flexion & rotation
Control group Group B (*n*=15)	NE	NE	NE	NE	NE	NE	Difference in 5% of all cases	NE

Table 6.2 Human locomotion restriction or loss (NE: not existed, SLR: straight leg raise).

Regarding the gait cycle duration results were as follows: In group A, the ipsilateral side had a mean gait cycle of 1,42sec (±0,11sec) and the controlateral side had a mean gait cycle of 1,39sec (±0,08sec), 95% C.I (table 6.4). The gait cycle of group B had duration of 1,21sec (±0,073, average from both extremities, 95% C.I). Totally, the gait cycle duration was found 14,8% longer in group A (scoliosis patients) compared to group B, $p<0{,}05$ (table 6.5).

Moderate idiopathic scoliosis in early adults produces a number of dysfunctional deformities that affect the postural system and produce abnormalities to the body segments that are closely related with the proper body-weight distribution and locomotion as well. The posture of the patient's bodies was asymmetrical with an inclination left or right according to the type of scoliosis. The basic movements of the trunk were affected and eventually restricted or even lost due to different types of deformity which are directly or indirectly connected with the deformity of scoliosis [18]. The average Cobb's angle in group A was $29{,}4^0$ and plumb line declination was 1,2cm (Table 6.3). Characteristic deformities were the functional leg length difference and the obliquity of pelvis that participated in an asynchronous gait cycle that both extremities in group A presented.

Clinical data	Scoliosis patients Group A (*n*=20)			Control group mean Group B (*n*=15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
	Ipsilateral side of the trunk			Trunk			
Cobb's angle (^{0})	29,4 (20^0-34^0)	2,97	1,30	0,45	0,25	0,13	**<0,01**
Apical rotation (grades)	+1 (0/+-4)	NE	NE	NE	NE	NE	NE
Plumb line declination (cm)	1,2	0,34	0,15	0,14	0,083	0,042	**<0,01**

Table 6.3 The average Plumb line declination (cm), Cobb's angle (^{0}) and apical rotation (grades) in scoliosis group and in healthy group. Significant differences are typed in bold and are accepted for P value <0.05, NE not existed, Confidence interval (±C.I).

Body-weight distribution of the lower extremities was unevenly in group A (Table 6.4) and the mean difference between them was 1,495 kg (±0,205, Confidence Interval C.I 95%) which is greater compared to 0,77 kg (±0,224), p<0,05, the mean difference between the extremities in control group B. Mean leg length discrepancy in group A was 1,49 cm (±0,2cm) while in group B the difference was 0,55 cm (±0,131), p<0,05. The gait cycle from both extremities in group A was asynchronous and the phases of walking were not executed in a simultaneous manner amongst them. The gait cycle in scoliosis patients was increased compared to healthy people and amongst the extremities in group A the ipsilateral side had a mean gait cycle at 1,42sec (±0,11sec) and the controlateral side had a mean gait cycle at 1,39sec (±0,076 sec) while the average from both extremities (group B) had a mean gait cycle at 1,21 sec (±0,073), p<0,05. Also, the mean gait cycle difference between the lower extremities from group A was 0,153sec (±0,039sec) and it is greater from the mean difference resulted from group B, that was 0,02 (±0,003sec), p<0,05. The statistical differences found in scoliosis (group A) patients between ipsilateral and controlateral extremity (side to side comparison) were concerning: The hip joint in the ipsilateral side of the trunk that had 6,4cm (±0,99) mean sagittal displacement (x axis), higher than the mean sagittal displacement (forward / backward) of the hip joint in the controlateral side that had 4,48cm (±0,53), p<0,05. The knee joint in the ipsilateral side of the trunk that had 6,74cm (±0,89) mean sagittal displacement, higher than the mean sagittal displacement of the knee joint in the controlateral side that had 4,8cm (±0,35), p<0,05 and the ankle joint at the ipsilateral side of the trunk had 6,46cm (±0,66) mean sagittal displacement, higher than the mean sagittal displacement of the ankle joint in the controlateral side that had 4,5cm (±0,34), p<0,05.

The center of gravity (midway between hips, few cm ahead S_2) in the ipsilateral side of the trunk that had 6,47cm (±0,88) mean sagittal displacement, higher than the mean sagittal displacement of the center of gravity in the controlateral side that had 4,35cm (±0,47), p<0,05.

Linear 3D velocity and 3D acceleration was lesser in the ipsilateral extremity but wasn't reached the level of any significant statistical difference.

Clinical data	Scoliosis patients Group A (*n* =20)						
	Average		Standard Deviation		Confidence Interval(±)		*P* value
	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	
Body weight distribution (Kg)	30,1	32,51	5,6560	7,2111	3,5501	3,1603	NS
Leg length discrepancy (cm)	83,96	85,56	6,2387	6,4155	2,7342	2,8116	NS
Gait cycle (sec)	1,415	1,393	0,2488	0,1680	0,1091	0,0736	NS

Table 6.4 The average body-weight distribution, Leg length discrepancy (LLD) and gait cycle in patients with moderate idiopathic scoliosis and healthy subjects. NS not significant, i.e. P value >0.05

Clinical data	Scoliosis patients Group A (*n* =20)			Control group mean Group B (*n* =15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
	Ipsilateral and controlateral extremities difference			Lower Extremities difference (mean)			
Body weight distribution (Kg)	2,405	1,824937	0,799	0,22	0,443471157	0,224423246	**<0,01**
Leg length discrepancy (cm)	1,6	0,479556	0,210	0,4866667	0,258751582	0,130943961	**<0,01**
Gait cycle (sec)	0,022	0,08420526	0,037	0,0026667	0,00507093	0,002566195	**<0,01**

Table 6.5 The average body-weight distribution, LLD and gait cycle difference between ipsilateral and controlateral extremity in patients with moderate idiopathic scoliosis and healthy subjects. Control group B represented by an average value from lower extremities due to minimal differences found amongst lower extremities. Significant differences are typed in bold and are accepted for P value <0.05

Clinical data	Scoliosis patients Group A (*n* =20)			Control group mean Group B (*n* =15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
	Ipsilateral extremity			Lower Extremities (mean)			
Body weight distribution (Kg)	30,1	5,6560	3,5501	30,57	5,82	2,94	NS
Leg length discrepancy (cm)	83,96	6,2387	2,7342	85,02	6,50	3,29	NS
Gait cycle (sec)	1,42	0,2488	0,1091	1,21	0,14	0,073	**<0,05**
	Controlateral extremity			Lower Extremities (mean)			
Body weight distribution (Kg)	32,51	7,2111	3,1603	30,57	5,82	2,94	NS
Leg length discrepancy (cm)	85,56	6,4155	2,8116	85,02	6,50	3,29	NS
Gait cycle (sec)	1,39	0,1680	0,0736	1,21	0,14	0,073	**<0,02**

Table 6.6 The average body-weight distribution, LLD and gait cycle in patients with moderate idiopathic scoliosis and healthy subjects. Control group B represented by an average value from lower extremities.
Significant differences are typed in bold and are accepted for P value <0.05, NS not significant, i.e. P value >0.05.

Kinematic data	Scoliosis patients Group A (*n*=20)						
	Average		Standard Deviation		Confidence Interval(±)		*P* value
Hip Joint	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	
Displacement X axis (cm)	6,365	4,48	2,257916	1,210307	0,989557	0,530431	**<0,03**
Displacement Y axis (cm)	2,915	2,94	0,856108	0,852489	0,375199	0,373613	NS
Displacement Z axis (cm)	-6,11	-5,835	1,185394	0,927518	0,519512	0,406495	NS
Displacement 3D (cm)	2,455	2,14	0,778308	0,871417	0,341102	0,381909	NS
Velocity 3D (m/sec)	0,0566	0,0576	0,017733	0,024095	0,007772	0,01056	NS
Acceleration 3D (m/sec^2)	0,185	0,21945	0,073339	0,135828	0,032142	0,059528	NS

Table 6.7 Kinematic data: The linear displacement (cm) in sagittal plane (x axis), gravitational plane (y axis), and frontal plane (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the hip joints amongst the ipsilateral and the controlateral extremity in scoliosis patients. Significant differences are typed in bold and are accepted for P value <0.05

Kinematic data	Scoliosis patients Group A (*n*=20)						
	Average		Standard Deviation		Confidence Interval(±)		*P* value
Knee Joint	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	
Displacement X axis (cm)	6,7421	4,805	2,045677	0,78972	0,89654	0,346104	**<0,01**
Displacement Y axis (cm)	7,4665	8,1705	4,714639	3,435169	2,066244	1,505501	NS
Displacement Z axis (cm)	-5,635	-5,56	0,901037	0,730825	0,39489	0,320292	NS
Velocity 3D (m/sec)	0,4398	0,45565	0,065678	0,090679	0,028784	0,039741	NS
Acceleration 3D (m/sec^2)	1,7667	1,9097	0,523855	0,857731	0,229585	0,37591	NS

Table 6.8 Kinematic data: The linear displacement (cm) in sagittal plane (x axis), gravitational plane (y axis), and frontal plane (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the knee joints amongst the ipsilateral and the controlateral extremity in scoliosis patients. NS not significant, i.e. P value >0.05

Kinematic data	Scoliosis patients Group A (*n*=20)						
	Average		Standard Deviation		Confidence Interval(±)		*P* value
Ankle Joint	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	
Displacement X axis (cm)	6,46	4,445	1,498912	0,786381	0,656915	0,34464	**<0,01**
Displacement Y axis (cm)	4,005	4,04	0,496806	0,456992	0,217731	0,200282	NS
Displacement Z axis (cm)	-5,32	-5,165	0,814733	0,88334	0,357066	0,387134	NS
Velocity 3D (m/sec)	0,6723	0,7176	0,124648	0,164049	0,054628	0,071896	NS
Acceleration 3D (m/sec^2)	2,7043	2,9467	1,17324	1,571771	0,514186	0,688846	NS

Table 6.9 Kinematic data: The linear displacement (cm) in sagittal plane (x axis), gravitational plane (y axis), and frontal plane (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the ankle joints amongst the ipsilateral and the controlateral extremity in scoliosis patients. Significant differences are typed in bold and are accepted for P value <0.05.

Kinematic data	Scoliosis patients Group A (*n*=20)						
	Average		Standard Deviation		Confidence Interval(±)		*P* value
Center of gravity	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	
Displacement X axis (cm)	6,465	4,33	1,9982295	1,075615	0,875747	0,4714	**<0,01**
Displacement Y axis (cm)	5,5355	5,5245	4,574625	4,47451	2,004881	1,961004	NS
Displacement Z axis (cm)	-5,84	-5,595	0,919038	0,7598303	0,402779	0,3330042	NS
Displacement 3D (cm)	11,56	10,945	3,493814	1,531245	1,531203	0,671085	NS
Velocity 3D (m/sec)	0,2959	0,3122	0,081886	0,074586	0,035888	0,032688	NS
Acceleration 3D (m/sec^2)	1,047	1,1728	0,371145	0,591763	0,162659	0,259347	NS

Table 6.10 Kinematic data: The linear displacement (cm) in sagittal plane (x axis), gravitational plane (y axis), and frontal plane (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the center of gravity amongst the ipsilateral and the controlateral extremity in scoliosis patients. NS not significant, i.e. P value >0.05

Comparison of group A and group B showed statistical significant difference in the following parameters (C.I 95%):

Hip measurements showed that ipsilateral side (group A) had -6,11cm (±0,52) increased mean frontal displacement (medial / lateral) compared to an average from both hips (group B) that had -5,10 cm (±0,23) mean frontal displacement (z axis), p<0,05. Also, group A had reduced mean sagittal (x axis-forward / backward direction), and increased frontal displacement in the hip joint of the controlateral side compared to an average value that both hips in group B produced (x axis, controlateral side, 4,48 cm, C.I ±0,53 vs. healthy extremities 6,41 cm, ±0.52 / z axis, controlateral side, -5,8 cm, ±0,41 vs. healthy ext. -5,10 cm, ±0,23), p<0,05. Relative to the knee joint, mean sagittal in the controlateral side in group A was lesser 4,48cm (±0,35) vs. 6,53cm (±0,43) compared to an average that both knees in group B produced . Knee's mean frontal displacement and mean vertical displacement in group A was increased in controlateral side comparison to group B average value from both extremities (controlateral, y axis 8,17cm, C.I ±1,51, z axis -5,56cm, ±0,32/ average value from both knees, group B, y axis 4,87cm,C.I ±0,64, z axis -4,73cm, ±0,20).

As for the knee joint in the ipsilateral side, the mean frontal displacement in scoliosis group was -5,6 cm (±0,39) and it is higher compared to -4,73 cm (±0,20) found in control group. The ankle's mean sagittal and frontal displacement in the ipsilateral side of the trunk (scoliosis patients) was higher compared to an average value resulted from both ankles in control group concerning x and z axis (ipsilateral, x axis 6,46cm, C.I ±0,66, z axis -5,32cm, ±0,36/ healthy extremities, x axis 4,74cm,C.I ±0,18, z axis -4,21cm, ±0,15), while mean frontal displacement was higher in the controlateral side of group A compared to group B (controlateral -5,17 cm, ±0,39 vs. healthy extremities -4,21 cm, ±0,15), p<0,05. The center of gravity had significantly reduced mean sagittal displacement in scoliosis patient's controlateral side, being 4,33cm (±0,47) for group A and 6,07cm (±0,52) for group B, p<0,05.

Linear 3D velocity and acceleration was lesser in Group A but wasn't reached the level of any significant statistical difference.

Kinematic data	Scoliosis patients Group A (*n*=20)			Control group mean Group B (*n*=15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
Hip Joint	Ipsilateral extremity			Lower extremities (mean)			
Displacement X axis (cm)	6,365	2,257916	0,989557	6,40666	1,031099	0,52180	NS
Displacement Y axis (cm)	2,915	0,856108	0,375199	2,69667	0,553388	0,280048	NS
Displacement Z axis (cm)	-6,11	1,185394	0,519512	-5,10333	0,451769	0,228622	**<0,02**
Displacement 3D (cm)	2,455	0,778308	0,341102	2,02	0,324478	0,164205	NS
Velocity 3D (m/sec)	0,0566	0,017733	0,007772	0,0646	0,004521	0,002288	NS
Acceleration 3D (m/sec^2)	0,185	0,073339	0,032142	0,233633	0,032479	0,016436	NS

Table 6.11 Kinematic data: The linear displacement (cm) in sagittal (x axis), gravitational (y axis), and frontal planes (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the hip joints amongst the ipsilateral extremity and the mean from both extremities in healthy people. Significant differences are typed in bold and are accepted for P value <0.05

Kinematic data	Scoliosis patients Group A (*n*=20)			Control group mean Group B (*n*=15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
Hip Joint	Controlateral extremity			Lower extremities (mean)			
Displacement X axis (cm)	4,48	1,210307	0,530431	6,40666	1,031099	0,52180	**<0,01**
Displacement Y axis (cm)	2,94	0,852489	0,373613	2,69667	0,553388	0,280048	NS
Displacement Z axis (cm)	-5,835	0,927518	0,406495	-5,10333	0,451769	0,228622	**<0,05**
Displacement 3D (cm)	2,14	0,871417	0,381909	2,02	0,324478	0,164205	NS
Velocity 3D (m/sec)	0,0576	0,024095	0,01056	0,0646	0,004521	0,002288	NS
Acceleration 3D (m/sec^2)	0,21945	0,135828	0,059528	0,233633	0,032479	0,016436	NS

Table 6.12 Kinematic data: The linear displacement (cm) in sagittal (x axis), gravitational (y axis), and frontal planes (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the hip joints amongst the controlateral extremity and the mean from both extremities in healthy people. NS not significant, i.e. P value >0.05.

Kinematic data	Scoliosis patients Group A (*n* =20)			Control group mean Group B (*n* =15)			
	Average	Standard Deviation	Confidence Interval(+-)	Average	Standard Deviation	Confidence Interval(+-)	*P* value
Knee Joint	Ipsilateral extremity			Lower extremities (mean)			
Displacement X axis (cm)	6,7421	2,045677	0,89654	6,5267	0,849762	0,43003	NS
Displacement Y axis (cm)	7,4665	4,714639	2,06624	4,866	1,272705	0,64406	**NS***
Displacement Z axis (cm)	-5,635	0,901034	0,39489	-4,7333	0,400743	0,2028	**<0,01**
Velocity 3D (m/sec)	0,4398	0,065678	0,028784	0,47	0,033434	0,01692	NS
Acceleration 3D (m/sec^2)	1,7667	0,523855	0,229585	1,938833	0,42825	0,21672	NS

Table 6.13 Kinematic data: The linear displacement (cm) in sagittal (x axis), gravitational (y axis), and frontal planes (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the knee joints amongst the ipsilateral extremity and the mean from both extremities in healthy people. Significantly different from the control group NS*

Kinematic data	Scoliosis patients Group A (*n* =20)			Control group mean Group B (*n* =15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
Knee Joint	Controlateral extremity			Lower extremities (mean)			
Displacement X axis (cm)	4,805	0,78972	0,346104	6,5267	0,849762	0,43003	**<0,01**
Displacement Y axis (cm)	8,1705	3,435169	1,505501	4,866	1,272705	0,64406	**<0,01**
Displacement Z axis (cm)	-5,56	0,730825	0,320292	-4,7333	0,400743	0,2028	**<0,01**
Velocity 3D (m/sec)	0,45565	0,090679	0,039741	0,47	0,033434	0,01692	NS
Acceleration 3D (m/sec^2)	1,9097	0,857731	0,37591	1,938833	0,42825	0,21672	NS

Table 6.14 Kinematic data: The linear displacement (cm) in sagittal plane (x axis), gravitational plane (y axis), and frontal plane (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the knee joints amongst the controlateral extremity and the mean from both extremities in healthy people.

Significant differences are typed in bold and are accepted for P value <0.05

Kinematic data	Scoliosis patients Group A (*n* =20)			Control group mean Group B (*n* =15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
Ankle Joint	Ipsilateral extremity			Lower extremities (mean)			
Displacement X axis (cm)	6,46	1,4989119	0,656915	4,7433	0,348909	0,1765695	**<0,01**
Displacement Y axis (cm)	4,005	0,496806	0,217731	3,84	0,350612	0,177431	NS
Displacement Z axis (cm)	-5,32	0,814733	0,357066	-4,21	0,299523	0,1515769	**<0,01**
Velocity 3D (m/sec)	0,6723	0,124648	0,054628	0,73667	0,080504	0,04074	NS
Acceleration 3D (m/sec^2)	2,7043	1,17324	0,514186	2,95912	0,53615	0,271324	NS

Table 6.15 Kinematic data: The linear displacement (cm) in sagittal plane (x axis), gravitational plane (y axis), and frontal plane (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the ankle joints amongst the ipsilateral extremity and the mean from both extremities in healthy people. Confidence interval (±C.I)

Kinematic data	Scoliosis patients Group A (*n* =20)			Control group mean Group B (*n* =15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
Ankle Joint	Controlateral extremity			Lower extremities (mean)			
Displacement X axis (cm)	4,445	0,786381	0,34464	4,7433	0,348909	0,1765695	NS
Displacement Y axis (cm)	4,04	0,456992	0,200282	3,84	0,350612	0,177431	NS
Displacement Z axis (cm)	-5,165	0,88334	0,387134	-4,21	0,299523	0,1515769	**<0,01**
Velocity 3D (m/sec)	0,7176	0,164049	0,071896	0,73667	0,080504	0,04074	NS
Acceleration 3D (m/sec^2)	2,9467	1,571771	0,688846	2,95912	0,53615	0,271324	NS

Table 6.16 Kinematic data: The linear displacement (cm) in sagittal plane (x axis), gravitational plane (y axis), and frontal plane (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the ankle joints amongst the controlateral extremity and the mean from both extremities in healthy people. NS not significant, i.e. P value >0.05

Kinematic data	Scoliosis patients Group A (*n*=20)			Control group mean Group B (*n*=15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
Center of gravity	Ipsilateral extremity			Lower extremities (mean)			
Displacement X axis (cm)	6,465	1,9982295	0,875747	6,07	1,0326457	0,52258123	NS
Displacement Y axis (cm)	5,5355	4,574625	2,004881	3,562	1,948334	0,985975	NS
Displacement Z axis (cm)	-5,84	0,919038	0,402779	-5,24333	0,4174184	0,21123896	NS*
Displacement 3D (cm)	11,56	3,493814	1,531203	10,37	1,307233	0,661539	NS
Velocity 3D (m/sec)	0,2959	0,081886	0,035888	0,348467	0,072243	0,036559	NS
Acceleration 3D (m/sec^2)	1,047	0,371145	0,162659	1,203533	0,379745	0,192174	NS

Table 6.17 Kinematic data: The linear displacement (cm) in sagittal (x axis), gravitational (y axis), and frontal planes (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the center of gravity amongst the ipsilateral extremity and the mean from both extremities in healthy people. Significant differences are typed in bold and are accepted for P value <0.05

Kinematic data	Scoliosis patients Group A (*n*=20)			Control group mean Group B (*n*=15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
Center of gravity	Controlateral extremity			Lower extremities (mean)			
Displacement X axis (cm)	4,33	1,075615	0,4714	6,07	1,0326457	0,52258123	**<0,01**
Displacement Y axis (cm)	5,5245	4,47451	1,961004	3,562	1,948334	0,985975	NS
Displacement Z axis (cm)	-5,595	0,7598303	0,3330042	-5,24333	0,4174184	0,21123896	NS
Displacement 3D (cm)	10,945	1,531245	0,671085	10,37	1,307233	0,661539	NS
Velocity 3D (m/sec)	0,3122	0,074586	0,032688	0,348467	0,072243	0,036559	NS
Acceleration 3D (m/sec^2)	1,1728	0,591763	0,259347	1,203533	0,379745	0,192174	NS

Table 6.18 Kinematic data: The linear displacement (cm) in sagittal (x axis), gravitational (y axis), and frontal planes (z axis) and the 3D linear velocity (m/sec) and acceleration (m/sec^2) exerted from the center of gravity amongst the controlateral extremity and the mean from both extremities in healthy people. NS not significant, i.e. P value >0.05

During the phases of gait cycle in group A, the angles of the knee joint (in sagittal axis-x axis) amongst ipsilateral and controlateral extremities did not showed any significant difference. Regarding knee range of motion scoliosis patients had seriously reduced range of angles (degrees) during gait cycle and a number of significant statistical differences were found amongst groups and included: in scoliosis group an initial contact (from double support phase and with heel strike) of the ipsilateral knee that was extended at 30,6^0 (±4,91), initial and mid swing phases with 26^0 (±2,18) and 50,7^0 (±2,31) flexion on average respectively, while the controlateral knee had 34,7^0 (±4,91) average extension at initial contact, 26,6^0 (±2,54) average flexion at initial swing phase and 51,2 (±4,93) average flexion in mid swing phase. In contrast group B (non-scoliosis group) had at initial contact an average extension at 2^0 (±0,51) and in initial and mid swing phases an average flexion at 41,5^0 (±0,42) and 74,5^0 (±0,43), respectively, p<0,05. The difference in the mean angular displacement of the knee joint, during the gait cycle, amongst the ipsilateral and the controlateral extremity in group A was not significant but it is higher ($1,44^0$,C.I ±1,07) compared to the mean angle difference exerted from the right and left knee in group B ($0,87^0$, ±0,09), p<0,05.

Kinematic data	Scoliosis patients Group A (*n* =20)						
	Average		Standard Deviation		Confidence Interval(±)		*P* value
Knee Joint (sagittal plane)	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	Ipsilateral extremity	Controlateral extremity	
Total angles of freedom	25,54	26,99	4,97	4,95	2,18	2,17	NS
Initial contact(^{0})	30,55	34,7	11,20	11,26	4,91	4,93	NS
Mid stance(^{0})	22,15	22,7	7,59	7,29	3,33	3,20	NS
Terminal stance(^{0})	6,45	6,71	5,42	4,90	2,38	2,15	NS
Initial swing(^{0})	25,9	26,6	6,08	5,80	2,66	2,54	NS
Mid swing(^{0})	50,65	51,2	5,26	7,19	2,31	3,15	NS
Terminal swing(^{0})	31,45	32,5	10,05	9,32	4,40	4,08	NS

Table 6.19 Kinematic data exerted from the knee joint in the sagittal plane (y axis) amongst the ipsilateral and the controlateral extremities in scoliosis patients during the phases of the gait cycle. NS not significant, i.e. P value >0.05

Kinematic data	Scoliosis patients Group A (*n* =20)			Control group mean Group B (*n* =15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
Knee Joint (sagittal)	Ipsilateral extremity			Lower Extremities (mean)			
Total Angles of freedom	25,54	4,97	2,18	31,27	0,88	0,45	**<0,01**
Initial contact(^{0})	30,55	11,20	4,91	2	1	0,51	**<0,01**
Mid stance(^{0})	22,15	7,59	3,33	24,5	0,80	0,41	NS
Terminal stance(^{0})	6,45	5,42	2,38	9,5	0,57	0,29	NS
Initial swing(^{0})	25,9	6,08	2,66	41,5	0,82	0,42	**<0,01**
Mid swing(^{0})	50,65	5,26	2,31	74,5	0,85	0,43	**<0,01**
Terminal swing(^{0})	31,45	10,05	4,40	29,5	1,15	0,58	NS
	Controlateral extremity			Lower Extremities (mean)			
Total Angles of freedom	26,99	4,95	2,17	31,27	0,88	0,45	**<0,02**
Initial contact(^{0})	34,7	11,26	4,93	2	1	0,51	**<0,01**
Mid stance(^{0})	22,7	7,29	3,20	24,5	0,80	0,41	NS
Terminal stance(^{0})	6,7	4,90	2,15	9,5	0,57	0,29	NS
Initial swing(^{0})	26,6	5,80	2,54	41,5	0,82	0,42	**<0,01**
Mid swing(^{0})	51,2	7,19	3,15	74,5	0,85	0,43	**<0,01**
Terminal swing(^{0})	32,5	9,32	4,08	29,5	1,15	0,58	NS

Table 6.20 Kinematic data exerted from the knee joint angles of freedom in the vertical plane (y axis) amongst: a) the ipsilateral extremity of scoliosis patients and the mean from both extremities in healthy people, b) controlateral extremity of scoliosis patients and the mean from both extremities in healthy people during the phases of the gait cycle. Significant differences are typed in bold and are accepted for P value <0.05, NS not significant, i.e. P value >0.05, Confidence interval (±C.I).

Kinematic data	Scoliosis patients Group A (*n* =20)			Control group Group B (*n* =15)			
	Average	Standard Deviation	Confidence Interval(±)	Average	Standard Deviation	Confidence Interval(±)	*P* value
Knee Joint (sagittal)	Ipsilateral and controlateral extremities difference			Lower extremities difference			
Total angles of freedom	1,44	2,44	1,07	0,87	0,18	0,09	**<0,01**
Initial contact(^{0})	4,15	7,17	3,14	1	0,16	0,08	**<0,03**
Mid stance(^{0})	0,55	2,06	0,90	0,49	0,22	0,11	**<0,01**
Terminal stance(^{0})	0,25	3,21	1,41	0,31	0,09	0,05	**<0,01**
Initial swing(^{0})	0,7	2,95	1,29	0,59	0,27	0,14	**<0,01**
Mid swing(^{0})	0,55	2,91	1,28	0,53	0,22	0,11	**<0,01**
Terminal swing(^{0})	1,05	6,97	3,06	0,71	0,67	0,34	**NS***

Table 6.21 The average differences of knee joint angles of freedom during phases of gait cycle between ipsilateral and controlateral extremities in patients with moderate idiopathic scoliosis and the lower extremities of healthy subjects. Significant differences are typed in bold and are accepted for P value <0.05, NS not significant, i.e. P value >0.05, NS* different from the control group but with no significance

CHAPTER 7
DISCUSSION

Young adults suffering from scoliosis, belong to a group of population with increased demands in everyday activities. The gait cycle plays important role in sport and occupational activities of people and can be analyzed with a simple and easy manner. The analyses could provide to us adequate information about the treatment plan of individuals with conditions affecting their ability to walk since MIS is the commonest type of scoliosis. We conducted this study to detect the effects of moderate idiopathic scoliosis on gait variables, of young adults, exerted from the hip the knee and ankle joint of the lower extremities as well as the center of gravity, and the correspondence of gait cycle relative to this kind of disorder, as compared to an able-bodied population and an asymmetric scoliosis posture.

The imbalance created by scoliosis affect the postural parameters of stability (center of mass and center of pressure) (Nault et al, 2002), the trunk (Raso et al, 1998), the coronal sacropelvic morphology (Mac Thiong et al, 2006) and thus an important determinant of gait that would be primarily affected (Della Croce et al, 2001) from this influence.

Studies showed that adolescent idiopathic scoliosis was not affecting the 3D displacement of pelvis during normal walking, resulted as a prolonged duration of activation of par vertebral muscles and equilibrium was maintained (Mahaudens et al, 2005) while other studies (Syczewska et al, 2006) showed that orientation of the pelvis during walking altered and this induces changes in gait stereotype.

Other studies showed that asymmetries in the gait pattern were detected in scoliosis patients and possible gait compensation is occurring, so that the subjects compensate on the controlateral pelvis / lower limb to that of the curve (Chockalingam et al, 2004). The IS patients generally produced higher sway area, lateral sway, sagittal sway, and sway radius than normal subjects. The cadence is smaller in the IS patients, but the stance phase and stride phase are similar to normal subjects (Chen et al, 1998). Other studies, (Mahaudens et al, 2009) suggested that patients with adult idiopathic scoliosis present no side to side differences but compared to healthy individuals a frontal pelvis, hip, and a transversal hip and sagittal knee motion restriction existed, the sagittal angular speed of the knee and ankle joint was decreased and the step length was reduced by 6 cm on average and the stance phase duration by only 2% on average. All these results indicated an almost physiological walk, even for those patients with severe scoliosis.

This study includes a major number of patients with thoraco-lumbar and lumbar primary curves because deformities at these levels are anatomically related to pelvis (Mac Thiong et al, 2006). From the kinesiology examination of scoliosis people in group A, it was clearly evident that an influence upon the axial musculoskeletal system existed and similar abbreviations noted in the study of Zabjek et al, 2005. Pain is possible an important factor that influence proper posture, according to previous studies (Cordover et al, 1997, Weiss, 1993) and locomotion as well as the pelvic obliquity secondary to scoliosis (Perry and Burnfield, 2010), the resultant leg length difference (White and Panjabi, 1990), and the body asymmetry which produce an asymmetrical body weight distribution on stance phase (Genthon et al, 2005).

The mean difference in the body weight distribution amongst the lower extremities in group A, compared to this from group B was higher as well as the discrepancy too. The gait cycle that produced by the lower extremities was affected and altered, and in group A was 14,8 % increased compared to the gait cycle presented in group B (p<0,05) which in terms was similar to optional gait cycle (Whittle et al,2002) as this shown in table 7.1. Also a higher mean difference existed, in the gait cycle, between ipsilateral and controlateral extremity. In contrast the mean difference amongst lower extremities in control group was minimal (p<0,05).

Approximate range (95 per cent limits) for general gait parameters in free-speed walking by normal FEMALE subjects of different ages				
Age (years)	Cadence (steps/min)	Cycle time (s)	Stride length (m)	Speed (m/sec)
13-14	103-150	0,80-1,17	0,99-1,55	0,90-1,62
15-17	100-144	0,83-1,20	1,03-1,57	0,92-1,64
18-49	98-138	0,87-1,22	1,06-1,58	0,94-1,66
50-64	97-137	0,88-1,24	1,04-1,56	0,91-1,63
65-80	96-136	0,88-1,25	0,94-1,46	0,80-1,52
Approximate range (95 per cent limits) for general gait parameters in free-speed walking by normal MALE subjects of different ages				
Age (years)	Cadence (steps/min)	Cycle time (s)	Stride length (m)	Speed (m/sec)
13-14	100-149	0,81-1,20	1,06-1,64	0,95-1,67
15-17	96-142	0,85-1,25	1,15-1,75	1,03-1,75
18-49	91-135	0,89-1,32	1,25-1,85	1,10-1,82
50-64	82-126	0,95-1,46	1,22-1,82	0,96-1,68
65-80	81-125	0,96-1,48	1,11-1,71	0,81-1,61

Table 7.1 General gait parameters in normal male and female subjects (reprinted from Whittle, 2002).

The time difference of the gait cycle amongst ipsilateral and controlateral extremity alters the phases of gait and their time of performance. When symptoms like tightness, elongation or shortening of the soft tissues that are surrounding the bony structures and pain was present, they were connected with both general gait attributes (unisommetry and unisochrony) in group A while the gait analysis in control group, with almost identical anthropometric characteristics, presented minimal differences in the physical quantities exerted from the major joints of the lower extremities and the center of gravity as well as in their timing of performance. The phases of gait were synchronized.

From the kinematic point of view, the motion restriction found in our study during the gait cycle and in a side-to-side comparison in group A, the mean sagittal (forward / backward) linear displacement of the hip, knee and ankle joints in the ipsilateral extremity (in the side of the convexity) was increased 29,6%, 25,7% and 33,2% respectively. The mean sagittal linear displacement concerning the center of gravity was increased 32,8% in the ipsilateral extremity. In contrast, studies marked no significant sagittal motion differences amongst the same joints of the lower extremities (Mahaudens et al, 2009) while other studies marked asymmetries but with the compensation to occur at the controlateral extremity (Chockalingam et al, 2004).

			Group A		Group B
			Ipsi lateral Ext.	Contro lateral Ext.	Right and Left Ext. average
Mean Linear Displacement	Hip	X Axis	6,37	4,48	6,41
		Y Axis	2,92	2,94	2,7
		Z Axis	-6,11	-5,84	-5,10
	Knee	X Axis	6,74	4,81	6,53
		Y Axis	7,47	8,17	4,87
		Z Axis	-5,64	-5,56	-4,73
	Ankle	X Axis	6,46	4,45	4,74
		Y Axis	4,01	4,04	3,84
		Z Axis	-5,32	-5,17	-4,21
	CoG	X Axis	6,47	4,33	6,07
		Y Axis	5,54	5,52	3,56
		Z Axis	-5,84	-5,60	-5,28

Table 7.2 Linear displacement of the major joints of the lower extremities and the center of gravity (CoG) compared between subjects with moderate idiopathic scoliosis and healthy people. (Ext / extremity).

Compared scoliosis patients with control group, in our study, the analysis showed that: the hip joint in the ipsilateral side (group A) had mean (z) frontal displacement (medial/lateral) increased 16,6%, while the hip joint in the controlateral side (group A) had mean (z) frontal displacement increaseded 12,6% and mean sagittal displacement decreased 30,1%. The knee joint in the ipsilateral side (group A) had mean (z) frontal displacement increased 19,1%. The knee joint in the controlateral side showed mean sagittal (x) displacement 26,5% decreased, the mean frontal displacement and the mean vertical (y) displacement (upwards/downwards) in group A was increased 17,5% and 40,5% respectively. The ankle joint in the ipsilateral side had increased mean sagittal and frontal displacement, 36,2% and 26,4% respectively. The ankle joint in the controlateral side showed 22,8% increased mean frontal displacement. The center of gravity in the controlateral side (group A) had mean (z) sagittal displacement decreased 28,6%.
Linear 3D velocity and acceleration was lesser in Group A but wasn't reached the level of any significant statistical difference.

From the above mentioned, scoliosis group had an increased sagital displacement existed as for the 3 major joints of the ipsilateral extremity (shorter extremity) and the transmission of the center of gravity compared to the controlateral extremity. When compared both groups, the controlateral hip joint and the center of gravity had lesser sagittal displacement. Scoliosis group had as a part of the compensation or the imbalance distorted motion in all 3 axes concerning the controlateral knee joint compared to healthy people. As for the ankle joint, distorted was the sagittal and the frontal linear displacement in the ipsilateral side. Frontal motion (medial / lateral) was affected in both knee and ankle joints from the extremities. The sagittal motion was decreased in the controlateral knee and increased in the ipsilateral ankle while Mahaudens et al, 2009 marked higher sagittal knee motion in scoliosis people but regarding the sagittal motion of the ipsilateral ankle did not observe any significant difference. The lateral sway (medial / lateral) in the z axis was higher in both knee and ankle joints from the ipsilateral and the controlateral side of group A and confirmed with other studies (Chen et al, 1998). The same studies mentioned that the vertical displacement was increased but from our analyses only the controlateral knee joint showed increased gravitational displacement (upwards / downwards).

Other researchers (Kramers et al, 2004) noted that sagittal plane hip motion followed a physiological pattern during gait cycle and the most significant and marked asymmetry was seen in the transverse plane, denoted as a torsional offset of the upper trunk in relation to the symmetrically rotating pelvis

In our study, the knee joint degrees of freedom were estimated in sagittal axis. During the phases of gait cycle, performed from young adults with moderate idiopathic scoliosis, we didn't found any significant statistical differences amongst ipsilateral and controlateral extremities as well as control group too but in scoliosis patients the ipsilateral and controlateral extremity overall angular degree of freedom was lesser.

		Group A			Group B	
	Linear	Ipsi lateral Ext.	Contro lateral Ext.	Diff.	Right and Left Ext. average	Diff.
	Gait Cycle (sec)	1,419	1,394	0,15	1,209	0.02
Hip Joint	3D Vel (m/sec)	0,057	0,058		0,065	
	3D Acc (m/sec^2)	0,19	0,22		0,24	
Knee Joint	3D Vel (m/sec)	0,44	0,46		0,47	
	3D Acc (m/sec^2)	1,77	1,91		1,94	
Ankle Joint	3D Vel (m/sec)	0,67	0,72		0,74	
	3D Acc (m/sec^2)	2,70	2,94		2,96	
CoG	3D Vel (m/sec)	0,30	0,31		0,35	
	3D Acc (m/sec^2)	1,04	1,17		1,2	

Table 7.3 Linear velocity and acceleration of the major joints of the lower extremities and the center of gravity (CoG) between subjects with moderate idiopathic scoliosis and control group. The gait cycle difference can be seen. (Diff / difference, Ext / extremity, Vel / velocity and Acc / aceleration).

Regarding the mean angular differences exerted by the ipsilateral and controlateral knee joint, during the phases of gait cycle in group A, significant statistical differences were found with the exception of the mean difference from terminal swing phase, compared to the mean angles exerted from the lower extremities in control group. This status indicated how the knee joint was affected in scoliosis group.

In group A, the ipsilateral knee had at initial contact 93% lesser extension, initial and mid swing phases with 37% and 32% lesser flexion on average compared to healthy extremities in control group, while the controlateral knee showed 94 % lesser extension at initial contact, initial and mid swing phases with 36%

and 31% lesser flexion on average in comparison to group B. This can be explained as a shorter stride length in conjunction to a higher sway radius of the distal parts of the lower extremities due to the fact that the gait cycle was increased in period of time in scoliosis group but with no significant statistical difference in the mean velocity and mean acceleration compared to control group. Other studies (Chen et al, 1998) didn’t show differences in stance and stride phases amongst scoliosis patients and healthy people. With the exception of the controlateral hip joint, knee joint, center of gravity and the ipsilateral ankle joint, especially the sagittal motion in scoliosis group is almost identical with control group. This gave to us the picture of a compensatory walking which was relatively close to normal walking.

From the above mentioned, these statistical significant differences might proven to be helpful in evaluating and treating the gait cycle of young adults with moderate idiopathic scoliosis. The observations provided important information about posture and the corresponding locomotion in such patients and create a basis for further studies on biomechanics and clinical entities like athletic and occupational performance, sense fatigue and pain symptoms.

CHAPTER 8
CONCLUSION

Scoliosis patients exhibit significantly impaired quality of life and young adults with MIS consist a population group with increased occupational and sports activities and gait cycle is of great importance. Gait analysis is used to identify and treat individuals with conditions affecting their posture and in terms their ability to walk.

We conducted this study in an effort to identify the degree that MIS influences the physical quantities exerted from the lower extremities and the transition of the CoG during the gait cycle of young adults in comparison to the gait cycle of healthy people.

Regarding this topic there are a lot of studies to our knowledge, but this study focuses on direct linear transformation method for analyses of gait cycle and transmission of the centre of gravity during walking.

Scoliosis patients (MIS) with moderate scoliosis resulted in pelvic obliquity and mild leg length discrepancy showed that abnormal posture of the body is capable to induce changes in locomotion during gait cycle and alter their gait manner. Despite that state, a compensatory walking existed and it was relatively close to normal walking. Scoliosis patients had their body-weight distribution unevenly distibuted amongst the lower extremities and they accomplish the gait cycle slower in comparison to healthy people. The phases of gait cycle were asynchronous between ipsilateral and controlateral extremities in scoliosis people and asymmetries can be found concerning a reduced sagittal displacement of the ipsilateral hip, knee and ankle joints as well as the transition of the center of gravity related to the same controlateral anatomical points during gait cycle. Scoliosis patients group showed disturbances in the behavior of the major joints of the lower extremities and the center of gravity in comparison to healthy people suggesting some kind of deformity and stiffness due to scoliosis. Pathologies affecting the gait cycle like inadequate extension at initial contact phase and inadequate flexion at initial and mid swing phases were present in scoliosis group as well as excessive abduction or valgus / varus. These statistical significant differences might proven to be helpful in evaluating and treating the gait cycle of young adults with moderate idiopathic scoliosis. Further studies focusing on improving range of motion, where found restricted, and/or leg length correction by orthotics and investigate their impact on gait and performance would be of great value.The observations provided important information about posture and the corresponding locomotion in such patients and create a basis for expansion of knowledge upon biomechanics and clinical entities like athletic and occupational performance, sense fatigue and pain symptoms.

SUMMARY

<u>Introduction:</u> Scoliosis influences the optimal posture and locomotion of the human body. Lumbar and thoracolumbar scoliosis has possibly an even greater impact due to the close relation with the pelvic region, a major determinant of gait. The effect of moderate idiopathic scoliosis on the gait cycle of young adults is of major interesting.

<u>Discussion-Hypothesis and aims:</u> Young adults (20-40 years old) with MIS (lumbar and thoracolumbar primary curves) present kinematic modifications regarding the convex (ipsilateral extremity) or concave (controlateral extremity) side of the body as well as variations compared to non-scoliosis individuals during the gait cycle. Aim of this study was to indentify these variations on the physical quantities exerted from the major joints of the lower extremities as well as the center of gravity too and examine their significance in adult patients with MIS.

<u>Materials and methods:</u> A cohort of twenty young adult patients (group A, 12 females- 8 males, mean age 39,7 years) having MIS with mean Cobb's angle 29^o (24^o to 34^o) and a control group (B) of fifteen (8 females, 7 males) of healthy individuals were submitted in clinical examination and 3-D gait analysis. Direct linear transformation (DLT) was used for analysis of linear displacement on the three axes (x, y and z) as well as 3D velocity and acceleration. The anatomical sites evaluated were concerning hip joint (greater trochanter), knee joint (lateral condyle) and ankle joint (lateral malleolus). Additionally, the gait cycle as well as the knee range of motion was examined. and the transision of the center of gravity (CoG).

<u>Results</u>: Mid leg length discrepancy (1,2cm ±0, 2, C.I 95%) was evident in scoliosis patients. Body-weight distribution between lower extremities was unevenly distributed in group A, $p<0,05$. The gait cycle in scoliosis patients showed increased duration compared to non scoliosis group patients. The ipsilateral side (to the convex side) had a mean gait cycle 1,42sec (±0,11sec) and the controlateral side (to the concave side) had a mean gait cycle 1,39sec (±0,076 sec). Group B had a mean gait cycle at 1,21 sec (±0,073), significantly faster, $p<0,05$. Regarding side to side comparison of the lower extremities in group A the following outcomes were identified: Hip and CoG were found to have greater sagittal (forward / backward) displacement on the ipsilateral side (to scoliosis curve) compared to the controlateral by 29,6% and 32,8% respectively ($p<0,05$). Knee joint linear displacement in the ipsilateral side was 25,7 % (increased), regarding sagittal axis, $p<0,05$. Ankle joint linear displacement in the ipsilateral (convex) side showed 33,2 % (increased), regarding sagittal axis, $p<0,05$. When compared group A to group B the following differences found: The hip joint in the ipsilateral side demonstrated frontal displacement increased by 16,6% compared to group B, c) The hip joint in the controlateral side had frontal (medial / lateral) displacement increased by 12,6 % and sagittal displacement decreased by 30,1%, compared to group B ($p<0,05$), d) The CoG in the controlateral side had mean sagittal displacement decreased by 28,6%, $p<0,05$ while in the ipsilateral side the difference was not significant. The knee joint in the ipsilateral side (group A) had mean (z) frontal displacement 19,1 % (increased), $p<0,05$. The knee joint in the controlateral side had mean sagittal (x) displacement 26,5 % (decreased), the mean frontal displacement and the mean

vertical (y-upwards / downwards) displacement in group A were 17,5 % and 40,5 % (increased) respectively, p<0,05. The ankle joint in the ipsilateral side had mean sagittal and frontal displacement, 36,2 % and 26,4 % (increased) respectively, p<0,05. The ankle joint in the controlateral side had 22,8 % mean frontal displacement (increased), p<0,05. The knee range of motion in scoliosis patients was seriously reduced during gait cycle. The ipsilateral knee joint had at initial contact (heel strike) 93 % lesser extension. At initial and mid swing phases 37 % and 32 % lesser flexion, compared to control group, p<0,05. The controlateral knee had 94 % lesser extension at initial contact, p<0,05. A initial and mid swing phases 36 % and 31 % lesser flexion, p<0,05.

Conclusion: Studies showed that scoliosis patients (MIS) presented asymmetries in the gait pattern and possible gait compensation on the controlateral extremity. Produced higher sway area in all axes than normal subjects but the stance phase and stride phase are similar to normal subjects. A sagittal knee motion restriction and a step length reduction by 6 cm on average were shown. In this study of patients with MIS and mild LLD, the gait cycle had increased duration compared to healthy people. Asymmetries exerted amongst the ipsilateral and controlateral hip, knee and ankle joint as well as the CoG during the gait cycle. Also asymmetries found in comparison to healthy people suggesting some kind of deformity and stiffness due to scoliosis. Some of these asymmetries agree to other studies and while others not. Pathologies affecting the gait cycle phases like inadequate extension or flexion may be responsible for a shorter stride length in conjunction to a higher sway radius of the distal parts of the lower extremities. A compensatory walking which was relatively close to normal walking existed. These statistical significant differences might prove to be helpful in evaluating and treat the gait cycle and can create a basis for intervention as well as further studies on biomechanics and entities like athletic and occupational performance, sense fatigue and pain symptoms.

Keywords: Adult Idiopathic scoliosis, gait cycle, DLT method, lower extremities.

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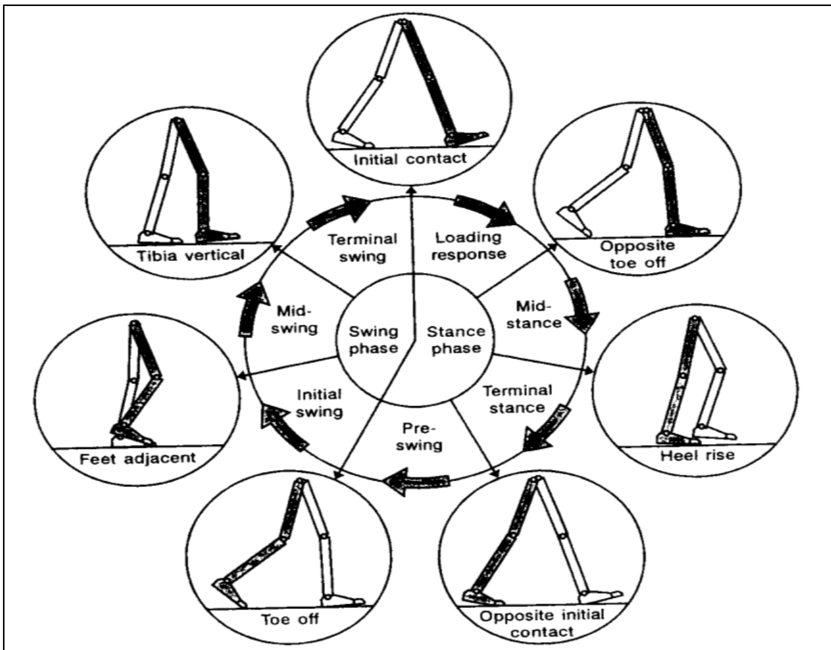
ABBREVIATIONS



Initial contact-IC
Opposite toe off-OTO
Heel rise-HR
Opposite initial contact-OIC
Toe off-TO
Feet adjacent-FA
Tibia vertical-TV
Loading response-LR
Mid-stance-MS
Terminal stance-TSt
Pre-swing-PSw
Initial swing-ISw
Mid-swing-MSw
Terminal swing-TSw
Base of support-BOS

(Adapted Whittle,2002)

APPENDIX

Absolute Reference System Definition

The Absolute Reference System is also called the Inertial Reference System or Global Reference System. *Definition:* A right-handed orthogonal triad <X, Y, Z > fixed in the ground. While a person is standing in an anatomically defined neutral posture, each of the axes is defined as:

+X axis pointing anteriorly (forward)

+Y axis pointing superiorly (upward) and in parallel with the field of gravity

+Z axis pointing rightward

Local Reference System Definition

Such a system is needed to describe segmental position and orientation with respect to the Absolute Reference System. Defined as a right-handed orthogonal triad $<X_{Li}, Y_{Li}, Z_{Li}>$ fixed to a point on the i^{th} body segment. An Absolute Reference System <**X, Y, Z** > is (where **T denotes the transport operator to the matrix)** first established so that the coordinates of these three markers are known. That is:

$$\boldsymbol{R}^1 = [x^1\ y^1\ z^1]^T \quad (1)$$
$$\boldsymbol{R}^2 = [x^2\ y^2\ z^2]^T \quad (2)$$
$$\boldsymbol{R}^3 = [x^3\ y^3\ z^3]^T \quad (3)$$

The Local Reference System $<X_{Li}, Y_{Li}, Z_{Li}>$ can be determines as:

$$\boldsymbol{R}^{1,2} = \boldsymbol{R}^1 - \boldsymbol{R}^2 \ (4) \quad , \quad \boldsymbol{R}^{1,3} = \boldsymbol{R}^1 - \boldsymbol{R}^3 \ (5) \quad ,$$

$$Z_{Li} = \boldsymbol{R}^{1,2} / |\boldsymbol{R}^{1,2}| \ (6) \quad , \quad X_{Li} = (\boldsymbol{R}^{1,3} \times \boldsymbol{R}^{1,2}) / |\boldsymbol{R}^{1,3} \times \boldsymbol{R}^{1,2}| \ (7) \quad \text{and} \quad Y_{Li} = Z_{Li} \times X_{Li} \ (8).$$

Segmental Reference System Definition

The Segmental Reference System is also called Anatomical Reference System and defined as a right-handed orthogonal triad $<X_{Si}, Y_{Si}, Z_{Si}>$ fixed to an anatomic point on the i^{th} body segment.

A free body segment is (e.g., a shank) moving in space with an Absolute Reference System <X, Y, Z>., and three landmark points a, b, and c are chosen from the segment: a is at medial malleolus, b is at lateral malleolus, and c is at tibial tubercle. Their **locations in the Absolute Reference System** are known as $\boldsymbol{R}^a$, $\boldsymbol{R}^b$, and $\boldsymbol{R}^c$. Based on these three position vectors we are ready to establish a segmental Reference System $<\mathbf{X}_{Si}, \mathbf{Y}_{Si}, \mathbf{Z}_{Si}>$ in the following way: $\boldsymbol{R}^{a,b} = \boldsymbol{R}^a - \boldsymbol{R}^b$ (9) , $\boldsymbol{R}^{a,c} = \boldsymbol{R}^a - \boldsymbol{R}^c$ (10)

$$Z_{Si} = \boldsymbol{R}^{a,b} / |\boldsymbol{R}^{a,b}| \ (11) \quad , \quad X_{Si} = (\boldsymbol{R}^{a,c} \times \boldsymbol{R}^{a,b}) / |\boldsymbol{R}^{a,c} \times \boldsymbol{R}^{a,b}| \ (12) \quad \text{and} \quad Y_{Si} = Z_{Si} \times X_{Si} \ (13).$$

Joint Reference System Definition

A Joint Reference System is a system that is fixed to a joint. It is needed in order to describe the relative movement of the body segments with respect to each other and defined as a right-handed triad $<\mathbf{X}_{Ji}, \mathbf{Y}_{Ji}, \mathbf{Z}_{Ji}>$ fixed to a point in the i^{th} joint that connects the i^{th} and the $(i + 1)^{th}$ segments. The axes are defined as:

$\mathbf{X}_{Ji}$ axis representing an axis of the $(1 + l)^{th}$ Segmental Reference System

$\mathbf{Z}_{Ji}$ axis representing an axis of the i^{th} Segmental Reference System

$\mathbf{Y}_{Ji}$ axis representing a floating axis that is the cross product of $\mathbf{Z}_{Ji}$ and $\mathbf{X}_{Ji}$.

Linear Transformation

Given an arbitrary point ***p*** on the segment, its position in the Absolute Reference System $\boldsymbol{R}^P$ and the Local Reference System $\boldsymbol{R}^P_{Li}$ can be described as:

$$\boldsymbol{R}^P = [x^P\ y^P\ z^P]^T \ (14) \qquad \text{and} \qquad \boldsymbol{R}^P_{Li} = [x^P_{Li}\ y^P_{Li}\ z^P_{Li}]^T \ (15)$$

Clearly, at time $t = t_O$, we have: $\boldsymbol{R}^P = \boldsymbol{R}^P_{Li}$ (16)

At time $t = t_n$ *the* segment, along with the Local Reference System, moves to another location which is represented by the position vector of O_{Li} ($\boldsymbol{R}^{OLi}$):

$$\boldsymbol{R}^{OLi} = [x^{OLi}\ y^{OLi}\ z^{OLi}]^T \quad (17)$$

Based on vector algebra, the position of point *p* that is described in the Local Reference System is related to the one in the Absolute Reference System in the following way:

$$\boldsymbol{R^P} = \boldsymbol{R^P}_{\boldsymbol{Li}} + \boldsymbol{R^{OLi}} \quad (18)$$

In order to generalize the description of the linear transformation, it is now necessary to introduce the quatrain or quaternion quantities method. A quatrain is a four-element vector including a spatial vector (three elements) and a scalar (one element). In general, the scalar is taken to be1. For example, the quatrain form of vector $\boldsymbol{R^P}$ is:

$$\boldsymbol{R^P} = [x^P\ y^P\ z^P\ 1]^{\mathrm{T}} \ (19)$$

We define a 4/4 matrix $\boldsymbol{Ş_{Li}}$ as: the linear transformation matrix from the Local Reference System *i* to the Absolute Reference System and is described in the following.

$$Ş_{Li} = \begin{pmatrix} 1 & 0 & 0 & x^{OLi} \\ 0 & 1 & 0 & y^{OLi} \\ 0 & 0 & 1 & z^{OLi} \\ 0 & 0 & 0 & 1 \end{pmatrix} \quad (20)$$

The linear transformation from Reference System II to Reference System I is:

$$\begin{pmatrix} x^P{}_I \\ y^P{}_I \\ z^P{}_I \\ 1 \end{pmatrix} = \begin{pmatrix} 1 & 0 & 0 & x_{\mathrm{I}}{}^{O\,\mathrm{II}} \\ 0 & 1 & 0 & y_{\mathrm{I}}{}^{O\,\mathrm{II}} \\ 0 & 0 & 1 & z_{\mathrm{I}}{}^{O\,\mathrm{II}} \\ 0 & 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} x^P{}_{\mathrm{II}} \\ y^P{}_{\mathrm{II}} \\ z^P{}_{\mathrm{II}} \\ 1 \end{pmatrix} = Ş_{\mathrm{I,\,II}} \begin{pmatrix} x^P{}_{\mathrm{II}} \\ y^P{}_{\mathrm{II}} \\ z^P{}_{\mathrm{II}} \\ 1 \end{pmatrix} \quad (21)$$

where $[x^k{}_j\ \ y^k{}_j\ \ z^k{}_j\ \ 1]^{\mathrm{T}}$ represents the coordinates of point *k* (*k* is either an arbitrary point *p* or the origin of the *J* Reference System O_J) in the *J* Reference System (J = I or II).

Displacement

Displacement describes the change in position of the body relative to a reference system. The change that is in a translational fashion is called linear displacement, whereas the change that is in a rotational fashion is called angular displacement

Velocity

The Velocity of a movement describes the speed of the change in position. The velocity is defined as the amount of change in position per unit time. If the position change is linear, the velocity is called the linear velocity. Likewise, if the position change is in rotation, the velocity is called angular velocity.

$$\boldsymbol{V^p}_{\boldsymbol{L}} = \boldsymbol{V^q}_{\boldsymbol{L}} + \boldsymbol{\Omega_L} \times \boldsymbol{R}^{\boldsymbol{p,q}}{}_{\boldsymbol{L}} \quad (22)$$

where $\boldsymbol{V^p}_{\boldsymbol{L}}$ and $\boldsymbol{V^q}_{\boldsymbol{L}}$ are the absolute linear velocity vectors at points *p* and *q* on one body segment, respectively: $\boldsymbol{\Omega_L}$ is the angular velocity vector of the body segment; and $\boldsymbol{R}^{\boldsymbol{p,q}}{}_{\boldsymbol{L}}$ is the position vector between these two points. The subscript *L* indicates the variable is expressed in the Local Reference System.

Acceleration

Acceleration describes the speed of change in velocity. It is defined as the change in velocity per unit time. As with displacement and velocity, there are linear and angular accelerations as well. Also, the linear acceleration is location dependent. The relation between the linear accelerations at two arbitrary points on a body segment is as follows:

$$\boldsymbol{a^p}_{\boldsymbol{L}} = \boldsymbol{a^q}_{\boldsymbol{L}} + \boldsymbol{a_L} \times \boldsymbol{R}^{\boldsymbol{p,q}}{}_{\boldsymbol{L}} + 2\,\boldsymbol{\Omega_L} \times \frac{\boldsymbol{d\,R}^{\boldsymbol{p,q}}{}_{\boldsymbol{L}}}{\boldsymbol{dt}} + \boldsymbol{\Omega_L} \times (\boldsymbol{\Omega_L} \times \boldsymbol{R}^{\boldsymbol{p,q}}{}_{\boldsymbol{L}}) \quad (23)$$

where $\boldsymbol{a^p}_{\boldsymbol{L}}$ and $\boldsymbol{a^q}_{\boldsymbol{L}}$ are the linear accelerations at points *p* and *q*, respectively, and $\boldsymbol{a_L}$ is the angular acceleration of the body segment.

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SUPPLEMENTARY

Our lab