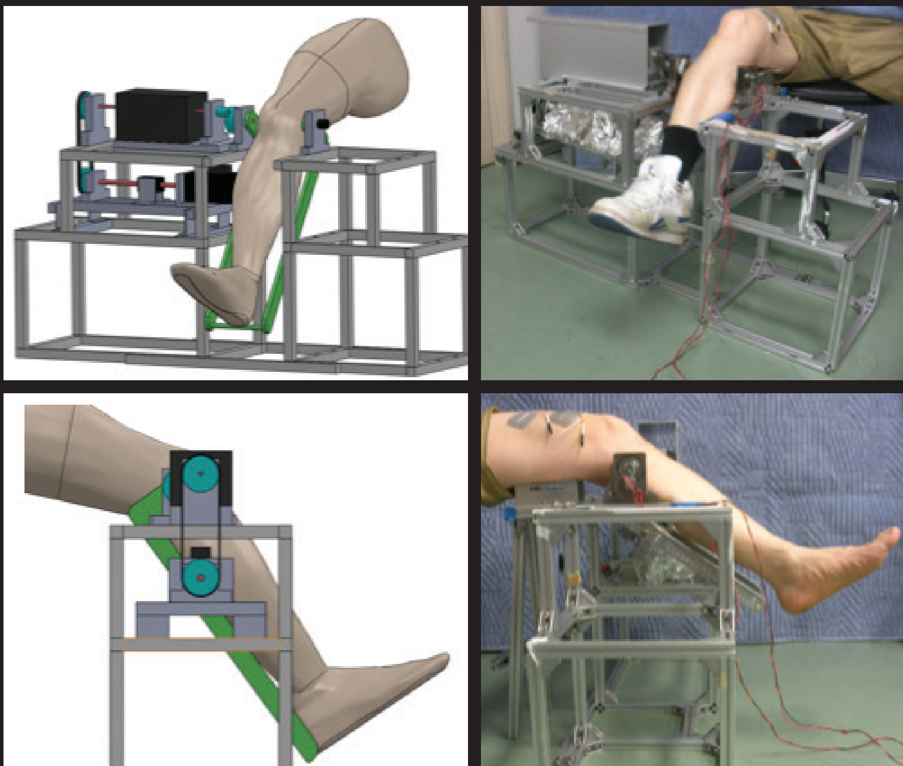
Advances in Therapeutic Engineering



Edited by

Wenwei Yu • Subhagata Chattopadhyay
Teik-Cheng Lim • U. Rajendra Acharya

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Preface

Therapeutic engineering (TE) is clearly focused to offer various technological assistance to curb the effects of malfunctioning of the human mind and body. It is exclusively a cross-disciplinary research area that mandates the understanding of the pathophysiology of an illness or a disorder and accordingly, mandates the design, development, and implementation of technology-based tools and software. The principal aim is to elevate the quality of life of the sufferers. Hence, there are several challenges in interfacing the clinical medicine and approapriate technology to offer the best possible solution. It is cutting-edge research and development in today's era of medical technology research.

This domain aims to implement the boons of technology in elevating the quality of life of patients who have developed either temporary or permanent damage of any part of their bodies. Numerous engineering algorithms are applied to develop, evaluate, and distribute technological solutions to problems confronted by individuals with disabilities. The book covers function areas including mobility, communications, hearing, vision, mental health and cognition, and diabetes management. The objective of this book is to cater various therapeutic processes and mechanisms applied to the field of healthcare in an integrated manner.

The human spine is one of the important and indispensable structures in the human body. Studies into the treatment of spinal diseases have played an important role in modern medicine. Many biomechanical models have been proposed to study dynamic behaviour as well as the biomechanics of the human spine and to develop new implants and new surgical strategies for treating these spinal diseases as discussed in Chapter 1.

Hand–eye coordination is a dynamic and effective measurement. Training of the hand–eye coordination of mentally ill patients is clearly a research challenge. Chapter 2 reviews the existing engineering methods and systems and then introduces the authors' collaborative research with the healthcare professionals at the Institute of Mental Health, Singapore, on psychomotor assessment and training.

Functional electrical stimulation (FES) is mainly used for restoring motor functions in people with disabilities due to stroke, head injury, diseases, and birth-related complications. Chapter 3 aims to study the central pattern generator model and its key mechanisms to design a control system for human lower limb movement control via FES. The corresponding simulation study demonstrates that the musculoskeletal system can perform smooth and accurate locomotion and maintain the posture well with this newly developed FES control system.

Chapter 4 discusses the recent functional magnetic resonance imaging (fMRI) study about neural plasticity. The development of a rehabilitation manoeuvre and the man–machine system may cause neural plasticity, and neuroimaging technology such as fMRI may be an effective means to inspect it.

Falls are the major causes of bone fractures in the elderly. Chapter 5 discusses an effective screening of the frail elderly with high fall risk. A novel device is developed for quantitative measurement of lower limb muscle strength and the authors attempt to verify its effectiveness. One of the outcomes of the study is to derive a threshold for lower limb muscle strength, which is critical for understanding the biomechanics of lower limbs.

Chapter 6 describes a novel actuator using a metal hydride (MH) alloy and its applications in rehabilitation and assistive devices. The MH actuator has several positive human-friendly properties, including a desirable force-to-weight ratio, a simple mechanism, noiseless, vibration-free drive, and inherent mechanical compliance, which has been improved over typical industrial actuators. These unique properties and their similarity to muscle actuation styles of expansion and contraction can be used for force devices for applications in human motion assist systems and rehabilitation exercise systems.

Amyotrophic lateral sclerosis (ALS) is a motor neuron degeneratative disorder resulting in the deterioration of voluntary motor function. In ALS subjects, the input device was designed to be operated by back-and-forth eye movement in an arbitrary direction to accommodate the limitations of eye movement. As ALS progresses, the ability to operate input devices deteriorates due to the gradual loss of voluntary motor function, loss of motivation, and physical pain. Chapter 7 discusses eye movement for the developed input device, as this motor function remains intact for a long period. The developed eye movement input device can be introduced irrespective of the disease stage, and it is designed for continuous use.

Advances in prosthetic devices in the last decade have improved the quality of life of those affected by limb loss. In the future, artificial limbs will be able to seamlessly integrate into the body of their users, emulating accurately their biological counterparts. Even though robotic technologies have brought considerable improvements to the field of cybernetics, in order to reach the goal of integration to the human body, future technologies need to provide mutual communication between the machine and the human body. The advances in the area to promote body awareness, such as sensory feedback and artificial skins, are discussed in Chapter 8.

FES is a technology to generate neural activity in an artificial way to activate muscles. The human response to FES is likely to be affected by several factors, such as spasticity, muscle fatigue, nerve habituation, and so forth. Chapter 9 discusses the development of a portable subthreshold stimulation and experiments to verify its effectiveness. The results show that the approach has enabled comparatively stable and durable function restoration assistance.

Chapter 10 reports a case of prosthetic hand rehabilitation using a brain–machine interface (BMI) and describes the development of the relevant technology. BMI technology shows promise for the rehabilitation of patients with serious paralytic impairments. The results of fMRI analysis of this BMI demonstrated brain adaptability with respect to prosthetic device use.

The induction heater–based treatment process is drawing interest in cancer cell destruction. Chapter 11 deals with the development of an induction heating unit for the above use in an efficient manner. The high frequency of the amplitude of the magnetic field intensity can produce unnecessary heating, which can lead to lesions surrounding healthy tissue via eddy current. This chapter discusses the parameters of alternating magnetic fields, which need to be chosen accurately.

Assistive technology (AT) covers all forms of aids and appliances that allow a differently abled individual to function in an optimum fashion in society. Chapter 12 covers a gamut of devices and adjustments that help the differently abled to function in society. All the models emphasise the role of a multidisciplinary approach to the design and conceptualisation of AT. The user must have an active part in the whole process for the AT to be meaningful and pleasurable to use.

Elderly and disabled people can benefit from the advancement of modern electronic devices, which can help them engage with the world. However, existing design practices often isolate elderly or disabled users by considering them users with special needs. Chapter 13 presents a brief description of the simulator, demonstrates its use through a couple of case studies, and presents its role in design optimisation and providing runtime adaptation to enhance the interaction experience among elderly users and people with disabilities.

Chapter 14 aims to describe a framework that resolves some of the semantic interoperability issues in e-health systems. This framework combines a dialogue game with a model of rational ontology mapping decision mechanisms to generate automatic dialogue between software agents, leading to resolution of certain types of ontological mismatches.

TE does not exclusively deal with tool/device development. Development of a decision support system is also a useful means of monitoring, screening, and managing diseases. The field of decision support systems is growing rapidly in many areas, including rehabilitation medicine. Chapter 15 provides a review of clinical decision support systems (CDSSs), particularly focusing on the mental health domain. It also focuses on the issue of screening and diagnoses of mental illnesses for prospective rehabilitation. Merits and demerits of several types of CDSSs are also discussed. Further, an attempt has been made to define an ideal CDSS from a rehabilitation perspective.

Chapter 16 discusses in detail the physiology of the human gait and its variations due to some degree of impairment. Comprehensive overviews of various research methodologies adopted to improve the impairment are explained. The concept of gait improvement through engineering-assistive devices has broadly been viewed under two segments: gait compensation and gait rehabilitation.

Cerebral palsy (CP) is an irreversible but nonprogressive motor condition characterised by abnormal control of movement or posture and muscle coordination. Electrical stimulation (ES) refers to the application of small electrical impulses to stimulate peripheral nerves and/or muscles to improve the impaired motor function. Chapter 17 suggests that ES has a favourable effect on gait and motor recovery in children with CP. The possible barriers for implementation, clinical implications, and important challenges for future research are highlighted.

A statistical pattern classification scheme for automated asthma detection using clinico–spirometric information is developed in Chapter 18. The clinical and spirometric information is captured by considering 42 features, out of which 19 (ranked) features are statistically significant ($p < .001$) in discriminating asthma and healthy groups. In this study, it is observed that the Bayesian classifier yielded 84.9% sensitivity, 97.7% specificity, and 94.17% overall accuracy.

In modern diagnostic scenarios, computer-aided diagnostic modules contribute enormously toward more accurate disease diagnosis. Chapter 19 proposes a malaria-screening module using textural characteristics of malaria-infected and noninfected erythrocytes. Their proposed approach predicts malaria with 96.73% accuracy, 99.72% sensitivity, and 84.39% specificity.

Chapter 20 reviews the various data mining techniques that have been developed and successfully implemented in healthcare management. The techniques covered include clustering, rule induction, artificial neural network, nearest neighbor, decision trees, and statistical techniques. The overlapping similarities and the differentiating factors among the various techniques are elucidated.

We are optimistic that the book will be very enlightening to readers of various disciplines and give them a flavour of recent research in the field of TE globally.

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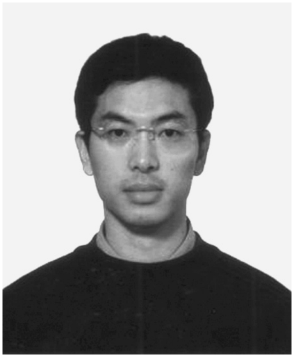

Wenwei Yu received his PhD in systems information engineering from Hokkaido University, Japan, in 1997 and served as an assistant professor in the System Information Engineering Department, School of Engineering, Hokkaido University from 1999 to 2003. He received his MD in rehabilitation medical science from Hokkaido University, Japan, in 2003. He was an exchange research fellow at the Center for Neuroscience, University of Alberta, Canada, in 2003, supported by the Researcher Exchange Program, Japanese Society for Promotion of Science (JSPS). Dr. Yu has been an associate professor in the Department of Medical System Engineering, School of Engineering, Chiba University, Japan, since April 2004 and a professor since September 2009. From July to October 2006, he worked in the AI Lab, Zurich University, Switzerland, as a visiting professor, supported by JSPS. He has authored and coauthored more than 200 papers in refereed journals, book chapters, and international and national conferences over the last 10 years. He is the reviewer for more than 10 international journals. He co-chaired the neural prosthetics and rehabilitation track of the 27th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC '05) held in September 2005 in Shanghai. His major interests are in neuroprosthetics, rehabilitation robotics, motor control, and biomedical signal processing. He is a member of IEEE, Robot Society of Japan, and Japanese Society for Medical and Biological Engineering.

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Teik-Cheng Lim obtained his PhD in engineering from the National University of Singapore (NUS) and is currently a head of programme at SIM University (UniSIM) as well as an accreditation member for an undergraduate degree programme in Singapore. He was awarded the Faculty of Engineering Annual Book Prize for his undergraduate studies and a research scholarship to pursue his doctorate at NUS. His research interests include auxetic materials, applied mechanics, and mathematical chemistry. He is the author of *An Introduction to Electrospinning and Nanofibers* (World Scientific, 2005) and editor for *Nanosensors: Theory and Applications in Theory, Healthcare and Defense* (Taylor & Francis, 2011), as well as lead author for more than 10 book chapters and more than 120 international journal papers. He is currently an editor-in-chief and editorial board member for six international journals and has served as a reviewer for about 30 international journals.

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Chapter 1

Development of a Human Spine Simulation System

Ian Gibson, Bhat Nikhil Jagdish, Gao Zhan, Khatereh Hajizedah, Hyunh Kim Tho, Huang Mengjie, and Chevanthie Dissanayake

Contents

1.1 Introduction to Spine Modelling

1.1.1 Introduction

The human spine is one of the important and indispensable structures in the human body. It undertakes many functions, most importantly, providing strength and support for the remainder of the human body, with particular attention to the heavy bones of the skull, as well as in permitting the body to move in ways such as bending, stretching, rotating, and leaning. Other functions include the protection of nerves, providing a base for the ribs, and offering a means of connecting the upper and lower body via the sacrum and pelvis. However, the human spine is also a very vulnerable part of our skeleton that is susceptible to many diseases and injuries, such as whiplash injury, low back pain, and scoliosis.

Whiplash injury to the human neck is a frequent consequence of rear-end automobile accidents and has been a significant public health problem for many years. Soft-tissue injuries to the cervical spine are basically defined as injuries in which bone fracture does not occur or is not readily apparent. A whiplash injury is therefore an injury to one or more of the many ligaments, intervertebral discs, facet joints, or muscles of the neck. Low back pain (LBP) is the most common disease compared to others and strongly associated with degeneration of intervertebral discs. LBP is usually seen in people with sedentary jobs who spend hours sitting in a chair in a relatively fixed position, with their lower back forced away from its natural lordotic curvature. This prolonged sitting causes health risks of the lumbar spine, especially for the three lower vertebrae, L3–L5. Eighty percent of people in the United States will have LBP at some point in their life.

Compared to LBP, scoliosis is a less common but more complicated spinal disorder. Scoliosis is a congenital three-dimensional deformity of the spine and trunk affecting between 1.5% and 3% of the population. In severe cases, surgical correction is required to straighten and stabilise the scoliosis curvature. Hence, studies into the treatment of these spinal diseases have played an important role in modern medicine. Many biomechanical models have been proposed to study dynamic behaviour as well as the biomechanics of the human spine, to develop new implants and new surgical strategies for treating these spinal diseases.

1.1.2 Motion of the Spine

A healthy spine provides the main support for the human body to allow movement in several planes. Motion of spine is usually measured in degrees of range of motion (ROM). The four movements measured are flexion, lateral flexion, extension, and rotation. The S-shape curve of a normal spine is able to absorb shock and maintain balance like a coiled spring to make sure of the full ROM. However, an abnormal curve of the spine, such as lordosis, kyphosis, and scoliosis, can lead to significant restrictions in spinal motion.

1.1.3 Vertebral Column

The spinal column (Figure 1.1) extends from the skull to the pelvis and is made up of 33 individual bones called vertebrae that are stacked on top of each other. The spinal column can be divided into five regions: 7 cervical vertebrae (C1–C7) in the neck, 12 thoracic vertebrae (T1–T12) in the upper back, 5 lumbar vertebrae (L1–L5) in the lower back, 5 bones (that are joined together in adults) to form the bony sacrum, and 3–5 bones fused together to form the coccyx or tailbone.

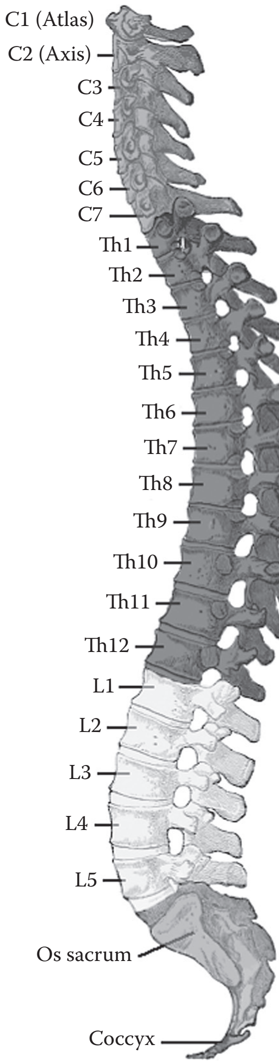

Figure 1.1 Spinal column.

1.1.4 Intervertebral Discs

The intervertebral discs (Figure 1.2) are soft-tissue structures situated between each of the 24 cervical, thoracic, and lumbar vertebrae of the spine. Their functions are to separate consecutive vertebral bodies. Once the vertebrae are separated, angular motions in the sagittal (forward and backward bending) and coronal planes (sideways bending) can occur.

1.1.5 Facet Joints

Facet joints are paired joints that are found in the posterior of the spinal column (Figure 1.3). Every vertebra has two facet joints to connect to the upper and lower vertebrae. The surfaces of each joint are covered by a cartilage that helps to smooth the movement between the two vertebrae. Certain motions are facilitated by these joints, such as bending forward, bending backward, and twisting. In addition, people can feel pain if the joints are damaged because of the connected nerves. Some experts believe that these joints are the most common reasons for spinal discomfort and pain.

1.1.6 Ligaments

The ligaments enable the spine to function in an upright position and the trunk to assume various positions for different activities. The spinal ligaments are extremely important for connecting the vertebrae and for keeping the spine stable. There are

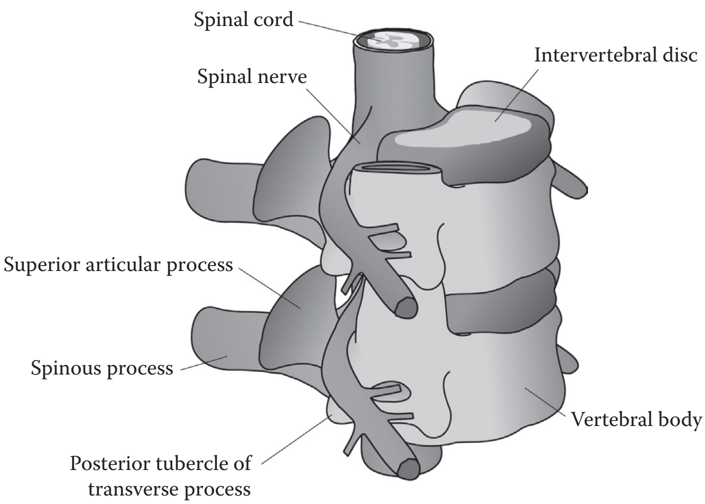

Figure 1.2 Intervertebral discs.

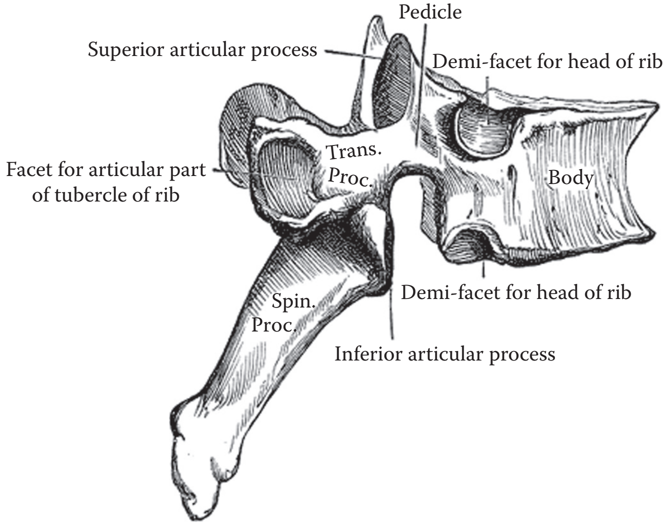

Figure 1.3 Facet joints of the spine.

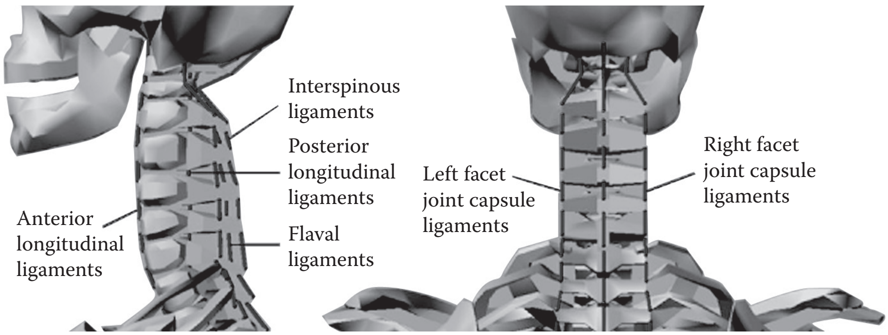

Figure 1.4 Ligaments of the spine.

various ligaments attached to the spine, with the most important being the anterior longitudinal ligament and the posterior longitudinal ligament (Figure 1.4), which runs from the skull all the way down to the base of the spine (the sacrum). In addition to the ligaments, there are also many muscles attached to the spine, which further help to keep it stable. The majority of the muscles are attached to the posterior elements of the spine.

1.1.7 Background on Computational Modelling Techniques

Models in biomechanics can be divided into four categories: physical models, *in vitro* models, *in vivo* models, and computer models. However, computer models have been extensively used due to their advantages over others, in that these models can provide information that cannot be easily obtained by other models, such as internal stresses or strains. They can also be used repeatedly for multiple experiments with uniform consistency, which lowers the experimental cost, and to simulate different situations easily and quickly. In computer models, multibody models (MBMs) and finite element models (FEMs) or a combination of the two are the most popular simulation tools that can contribute significantly to our insight of the biomechanics of the spine.

Although a great deal of computational power is required, FEMs are helpful in understanding the underlying mechanisms of injury and dysfunction, leading to improved prevention, diagnosis, and treatment of clinical spinal problems. These models often provide estimates of parameters that *in vivo* or *in vitro* experimental studies either cannot or are difficult to obtain accurately. Basically, FEMs are divided into two categories: models for dynamic study and models for static study. Models developed for static study generally are more detailed in representing spinal geometries. Although this type of model can predict internal stresses, strains, and other biomechanical properties under complex loading conditions, they generally only consist of one or two motion segments and do not provide more insight for the whole column. Models for dynamic study generally include a series of vertebrae (as rigid bodies) connected by ligaments and discs modelled as springs. These models could only locally predict the kinematic and dynamic responses of a certain part of the spine under load. In addition to static and dynamic investigations, FEMs have also been widely used for years to study scoliosis biomechanics. Thoroughly understanding the biomechanics of spine deformation will help surgeons to formulate treatment strategies for surgery as well as design and development of new medical devices involving the spine. Due to the complexity of spine deformities, FEMs of scoliotic spines are usually restricted to two-dimensional models or sufficiently simplified into three-dimensional elastic beam element models. Although these models show some promising preliminary results, extensive validation is necessary before using the models in clinical routine.

Compared to FEMs, MBMs have advantages such as less complexity, less demand on computational power, and relatively simpler validation requirements. MBMs possess the potential to simulate both the kinematics and kinetics of the human spine effectively. In MBMs, rigid bodies are interconnected by bushing elements, pin (two-dimensional) and/or ball-and-socket (three-dimensional) joints. MBMs can also include many anatomical details while being computationally efficient. In these models, the head and the vertebrae are modelled as rigid bodies, and soft tissues (intervertebral discs, facet joints, ligaments, and muscles) are usually modelled as massless spring-damper elements. Such MBMs are capable of producing biofidelic responses. Generally, MBMs can be broken down into two categories: car collisions and whole-body vibration investigations. In the former, displacements of the head with respect to the torso, accelerations, intervertebral motions, and neck forces/moments can provide good predictions for whiplash injury. In the latter, MBMs are helpful for determining the forces acting on the intervertebral discs and end plates of lumbar vertebrae. In both cases, MBMs are only focused either on the cervical spine or on the lumbar spine. Since these spine segments are partially modelled in detail, it is impossible to investigate the kinematics of the thoracic spine region. In other words, global biodynamic response of the whole spine has still not been studied thoroughly.

1.2 Development of a Spine Simulation System

1.2.1 Introduction to LifeMOD™ Simulation Software

Recently, many software applications have been developed for impact simulation, ergonomics, comfort study, biomechanical analysis, movement simulation, and surgical planning. Such software enables users to perform human body modelling and interaction with the environment where the human motion and muscle forces can be simulated. These tools are very useful for simulating the human–machine behaviour simultaneously. LifeMOD™ from Biomechanics Research Group is a leading simulation tool that has been designed for this purpose.

The LifeMOD™ Biomechanics Modeller is a plug-in module to the ADAMS (Automatic Dynamic Analysis of Mechanical Systems) physics engine, produced by MSC Software Corporation to perform multibody analysis. It provides a default MBM of the skeletal system that can be modified by changing anthropometric sizes such as gender, age, height, and weight. The created human body may be combined with any type of physical environment or system for full dynamic interaction. The results of the simulation are human motion, internal forces exerted by soft tissues (muscles, ligaments, and joints), and contact forces at the desired location of the human body. Full information on the LifeMOD™ Biomechanics Modeller can be found online [1].

In Sections 1.2.2 to 1.2.7, the development process of a discretised musculoskeletal spine model is presented thoroughly. This process includes five main stages: generating a default human body model, discretising the default spine segments, implementing ligamentous soft tissues, implementing lumbar back muscles, and adding intra-abdominal pressure.

1.2.2 Generating a Default Human Body Model

The usual procedure for generating a human model is to create a set of body segments, followed by redefining the fidelity of the individual segments. The body segments of a complete standard skeletal model are first generated by LifeMOD™ depending on the user's anthropometric input. The model used in this study was a median model with a height of 1.78 m and a weight of 70 kg created from the internal GeBod anthropometric database. By default, LifeMOD™ generates 19 body segments represented by ellipsoids. Then, some kinematic joints and muscles are generated for the human model. Figure 1.5 shows the base model in this study.

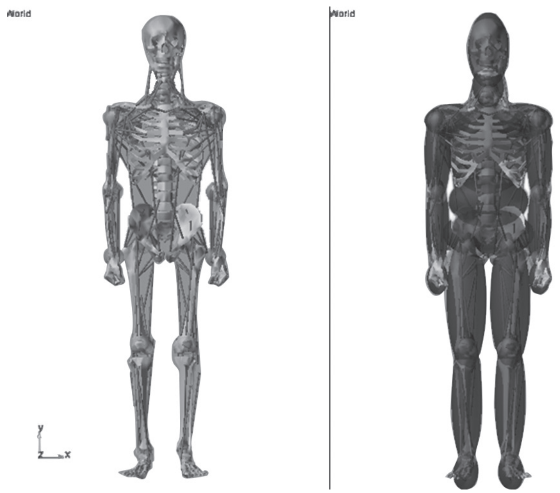

Figure 1.5 Base model for study.

1.2.3 Discretising the Default Spine Segments

To achieve a more detailed spine model, the improvement of the default spine model mentioned in Section 1.2.1 is required and can be done in the three following steps: refining the spine segments, reassigning muscle attachments, and creating the spinal joints.

1.2.3.1 Refining the Spine Segments

From the base human model, the segments may be broken down into individual bones for greater model fidelity. Every bone in the human body is included in the generated skeletal model as a shell model. To discretise the spine region, the standard ellipsoidal segments representing the cervical (C1–C7), thoracic (T1–T12), and lumbar (L1–L5) vertebral groups are firstly removed. Based on input such as the centre of mass location and the orientation of each vertebra, the individual vertebra segment is then created. Figure 1.6 shows all ellipsoidal segments of 24 vertebrae in the cervical, thoracic, and lumbar regions after discretising.

1.2.3.2 Reassigning Muscle Attachments

The muscles are attached to the respective bones based on geometric landmarks on the bone graphics. With the new vertebra segments created, the muscle attachments to the original segment must be reassigned to be more specific to the newly created vertebra segments. The physical attachment locations will remain the same. Figure 1.7a and b shows the anterior and posterior view of several muscles in neck/trunk regions. Table 1.1 lists the attachment locations of these muscles.

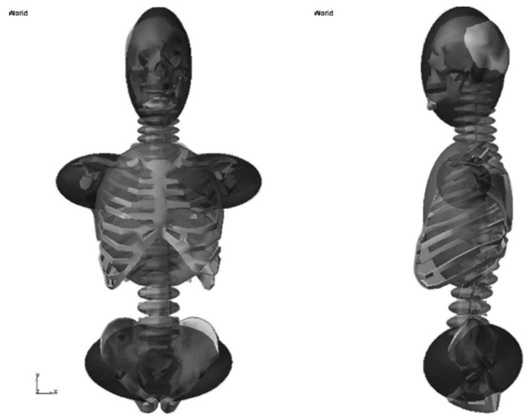

Figure 1.6 Ellipsoid segments of all discretised vertebrae.

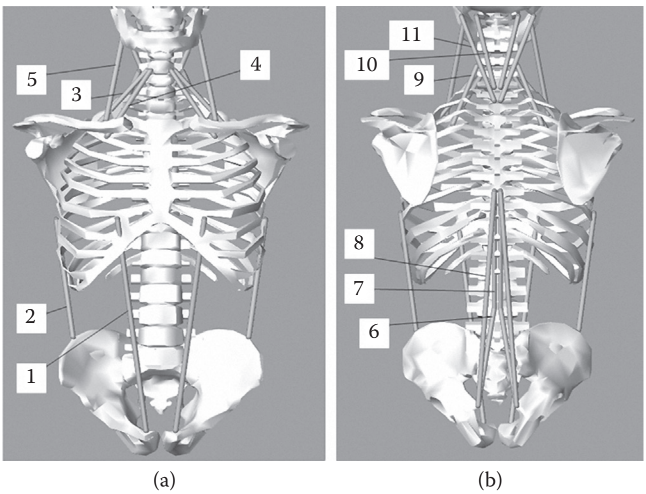

Figure 1.7 Muscles in neck and trunk regions: (a) anterior and (b) posterior.

1.2.3.3 Creating the Spinal Joints

It is necessary to create individual nonstandard joints representing intervertebral discs between newly created vertebrae. The spinal joints are modelled as torsional spring forces, and the passive six degrees of freedom's (DOFs) jointed action can be defined with user-specified stiffness, damping, angular limits, and limiting stiffness values. These joints are used in an inverse dynamics analysis to record the joint angulations while the model is being simulated. The properties of the joints can be found in the literature [2–5]. Figure 1.8 shows spinal joints representing intervertebral discs.

Table 1.1 Attachment Locations of Neck and Trunk Muscle Set

Index	*Muscle*	*Attach Proximal*	*Attach Distal*
1	Rectus abdominis	Sternum	Pelvis
2	Obliquus externus	Ribs	Pelvis
3	Scalenus medius	C5	Ribs
4	Scalenus anterior	C5	Ribs
5	Sternocleidomastoideus	Head	Scapula
6	Erector spinae 2	L2	Pelvis
7	Erector spinae 3	T7	L2
8	Erector spinae 1	T7	Pelvis
9	Scalenus posterior	C5	Ribs
10	Splenius cervicis	Head	C7
11	Splenius capitis	Head	T1

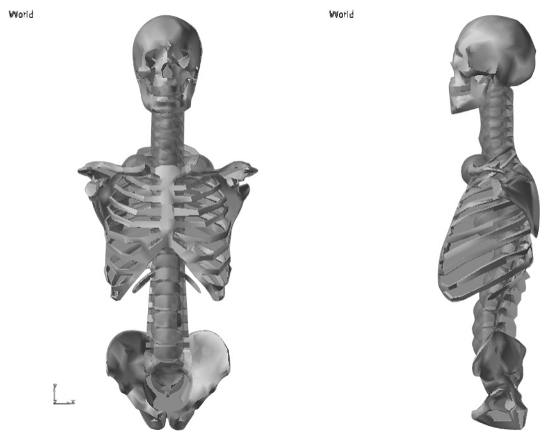

Figure 1.8 Spinal joints representing intervertebral discs.

1.2.4 Creating the Ligamentous Soft Tissues

To stabilise the spine model, interspinous, flaval, anterior longitudinal, posterior longitudinal, and capsule ligaments are created. Figure 1.4 displays various types of ligaments attached to vertebrae in the cervical spine region.

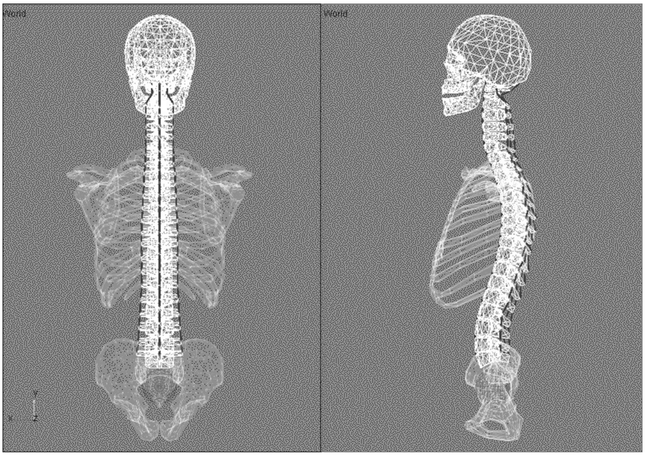

Figure 1.9 Ligaments attached to the whole spine.

Figure 1.9 shows side and rear views of all ligaments of the whole spine running from the skull down to the pelvis. These ligaments surrounding the spine will guide segmental motion and contribute to the intrinsic stability of the spine by limiting excessive motion. The stiffness of these ligaments is referenced in [6,7].

1.2.5 Implementing Lumbar Muscles

1.2.5.1 Multifidus Muscle

The multifidus muscle is divided into 19 fascicles on each side according to descriptions by Bogduk and colleagues [8,9]. The multifidus can be modelled as three layers, with the deepest layer having the shortest fibres and spanning one vertebra. The second layer spans over two vertebrae, while the third layer goes all the way from L1 and L2 to posterior superior iliac spine [10]. The rather short span of the multifidus fascicles makes it possible to model them as line elements without via-points (Figure 1.10a).

1.2.5.2 Erector Spinae Muscle

According to descriptions by Macintosh and Bogduk [11,12], there are four divisions of the erector spinae: longissimus thoracis pars lumborum, iliocostalis lumborum pars lumborum, longissimus thoracis pars thoracis, and iliocostalis lumborum pars thoracis. The fascicles of the longissimus thoracis pars lumborum and iliocostalis lumborum pars lumborum originate from the transverse processes of the lumbar vertebrae and insert on the iliac crest close to the posterior superior iliac spine [10]. The fascicles of the longissimus thoracis pars thoracis originate

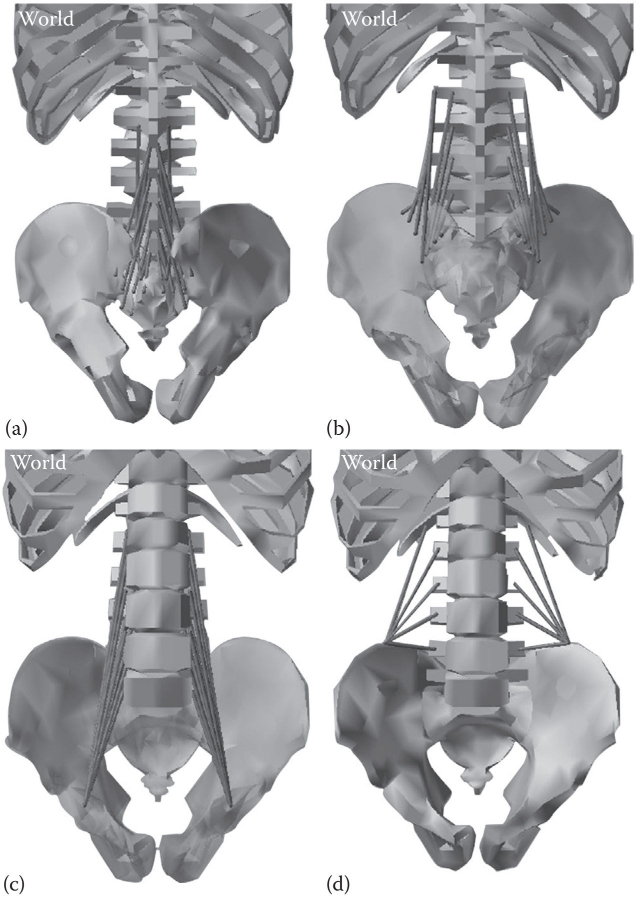

Figure 1.10 Muscle attachments in the lower spine and sacrum.

from the costae 1–12 close to the vertebrae and insert on the spinous process of L1 down to S4 and on the sacrum. The fascicles of the iliocostalis lumborum pars thoracis originate from the costae 5–12 and insert on the iliac crest. Since muscles of the two pars thoracis are automatically generated by LifeMOD™, only muscles of the two pars lumborum need to be added to our model, as shown in Figure 1.10b.

1.2.5.3 Psoas Major Muscle

The psoas major muscle is divided into 11 fascicles according to different literature sources [13–15]. The fascicles originate in a systematic way from the lumbar vertebral bodies and T12 and insert into the lesser trochanter minor of the femur with a via-point on the pelvis (iliopubic eminence) (Figure 1.10c). Bogduk found that the psoas major had no substantial role as a flexor or extensor of the lumbar spine, but rather that the psoas major exerted large compression and shear loading on the lumbar joints [13]. This implies that the moment arm for the flexion/extension direction is small and therefore the via-points for the path were chosen in such a way that the muscle path ran close to the centre of rotation in the sagittal plane.

1.2.5.4 Quadratus Lumborum Muscle

For modelling the quadratus lumborum, the description given by Stokes and Gardner-Morse was followed [16]. They proposed to represent this muscle by five fascicles. The muscle originates from costa 12 and the anterior side of the spinous processes of the lumbar vertebrae and has in the model a common insertion on the iliac crest (Figure 1.10d).

1.2.5.5 Abdominal Muscles

Two abdominal muscles are included in the model: obliquus externus and obliquus internus. Modelling of these muscles requires the definition of an artificial segment with a zero mass and inertia [10]. This artificial segment mimics the function of the rectus sheath on which the abdominal muscles can attach (Figure 1.11a). The obliquus externus and the obliquus internus are divided into six fascicles each [16]. Two of the modelled fascicles of the obliquus externus run from the costae to the iliac crest on the pelvis, while the other four originate on the costae and insert into the artificial rectus sheath as can be seen in Figure 1.11a. Three of the modelled fascicles of the obliquus internus run from the costae to the iliac crest, while the other three originate from the iliac crest and insert into the artificial rectus sheath (Figure 1.11b).

1.2.6 Adding Intra-Abdominal Pressure

Since LifeMOD™ and ADAMS provide tools that only generate concentrated or distributed forces, it is not possible to directly implement intra-abdominal pressure into the spine model. To overcome this difficulty, a new approach to intra-abdominal pressure modelling is proposed. Initially, an equivalent spring structure able to mimic all mechanical properties of intra-abdominal pressure such as tension/compression, anterior/posterior shear, lateral shear, flexion/extension, lateral bending, and torsion is created (Figure 1.12). After that, the translational and torsional stiffnesses of the spring structure are determined. Finally, since adding this spring structure into the spine model is quite troublesome, a bushing element that can specify all stiffness properties of the structure is used instead (Figure 1.13).

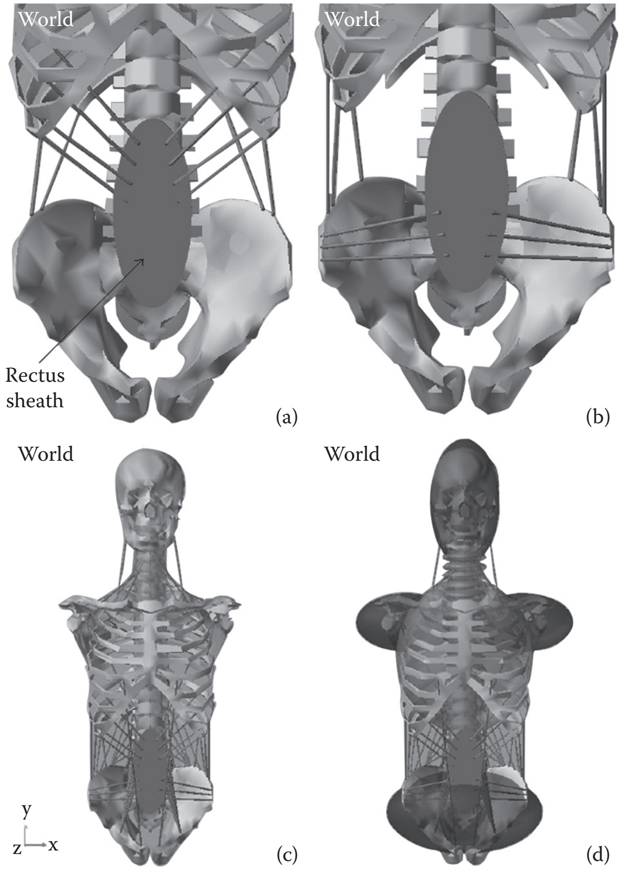

Figure 1.11 Abdominal muscle attachments.

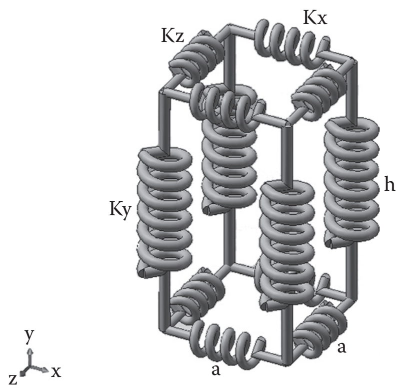

Figure 1.12 Equivalent spring structure to mimic intra-abdominal pressure.

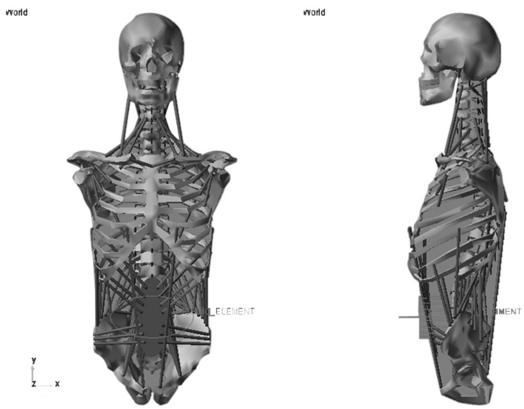

Figure 1.13 Intra-abdominal pressure modeled using bushing element.

1.2.7 Validation of the Detailed Spine Model

To validate the detailed spine model presented in Sections 1.2.2 through 1.2.6, two approaches are used and presented as follows:

- With the same extension moment generated in upright position, the axial and shear forces in the L5–S1 disc calculated in the model are compared to those obtained from Zee's model [10] and experimental data [17].
- While a subject holds a crate of beer weighing 19.8 kg, the axial force of the L4–L5 disc is computed and compared with *in vivo* intradiscal pressure measurements [18].

In the first approach, a gradually increasing horizontal force was applied onto the vertebra T7 of the spine model from posterior to anterior in the sagittal plane. From this force, axial and shear forces as well as the moment about the L5–S1 disc were calculated. Zee's model estimated an axial force of 4520 N and shear force of 639 N in the L5–S1 disc at a maximum extension moment of 238 Nm. Meanwhile, to obtain the same extension moment, the external force that needs to be applied in the present model is 1260 N. Corresponding with this force, the axial and shear forces obtained in the model were 4582 N and 625 N respectively. This is in accordance with the results presented by McGill and Norman [17] who found axial forces in the range of 3929–4688 N and shear forces up to 650 N.

In the second approach, a comparison was made with *in vivo* intradiscal pressure measurements of the L4–L5 disc as reported by Wilke et al. [18]. They measured a pressure of 1.8 MPa in the L4–L5 disc while the subject (body mass: 70 kg; body height: 1.74 m) was holding a full crate of beer (19.8 kg) 60 cm away from the chest. The disc area was 18 cm^2, and based on this, the axial force was calculated to be 3240 N. The same situation was simulated using the spine model in our research. The estimated axial force was 3161.6 N. This is a good match considering the fact that no attempt was made to scale the model to the subject in this study. The body mass and body height of the subject in this study are quite similar to the body mass and height used in the model.

1.3 Applications of a Human Spine Simulation System

The entirely discretised multibody spine model in our study can be used in numerous medical applications, such as product design, clinical treatment, and surgical training. In this section, some preliminary results based on this spine model will be presented.

1.3.1 Developing a Human–Chair Interface to Provide Means of Designing Effective Seating Solutions

LBP is a complex condition and is not entirely understood even today. The source of the pain can be attributed to factors such as muscular dysfunction, joint irritation, breakdown of vertebral bodies, postural distortions, and severe spinal deformities such as scoliosis [19]. In complicated medical conditions, emphasis is placed on controlling the risk factors involved. In the case of LBP, optimising the spine's position to resist the compressive forces of gravity is a logical place at which to address the high risk of fracture or change due to stresses [20]. Sitting increases disc pressure [21], and therefore, there has been a consensus that seating is a contributing factor to the risk of LBP. Thus, understanding sitting posture, sitting behaviour, and the corresponding force variations in the spine can assist in treating LBP.

Virtual platforms are now being used to investigate the effects of implants, understand gait cycles, simulate sports actions, study injury scenarios, and offer solutions that may reduce undue stresses and strains on the musculoskeletal system. The ergonomics of sitting have been studied in healthy individuals and for LBP; remedies have been suggested by modifications to chair design or to posture (apart from medical treatment). Wheelchair-bound patients often sit for extended periods of time in a fixed position, which contributes to the development of LBP. However, in wheelchair seating, changing the seat or posture is not an easy option. Firstly, wheelchairs are expensive and are usually not changed unless the patient's needs drastically change or he/she outgrows it. Also, many patients cannot perform dynamic seating actions (small shifts in position to momentarily relieve pressures) or change posture. Currently, physiotherapists and occupational therapists rely on personal experience in order to determine how and what parts of the body to support in the patient. Today, projects such as this are attempting to assist in identifying the major stresses in the spine when seated in order to develop a system whereby once a patient's data is imported, the major stresses in the spine can be determined and the best places for support identified so that the risk of conditions such as LBP can be reduced.

1.3.1.1 Using MOCAP to Capture Seating Motion

To study seating action as it is performed in real life, Vicon Motion Capture (MOCAP) system (see Figure 1.14) was used to capture the seating motion and analyse the interaction of the spine with the external environment (chair). Vicon Nexus was the first life science–specific MOCAP software available in the market. It is a validated system that is being used for gait analysis and rehabilitation, posture, balance and motor control, sports performance, and others.

The technology underpinning the Vicon camera systems are based on small retroreflective markers attached to specific places on the subjects' body. The Vicon cameras emit strobe light, which is reflected back into the cameras from the markers, giving a clear, greyscale view of each marker. The location coordinates of each marker are then calculated from the greyscale image and forwarded to a computer. This information, received from all the cameras, establishes highly accurate three-dimensional trajectories. The Vicon Nexus and Vicon BodyBuilder software assist in analysing the results from the camera system.

1.3.1.2 Data Preparation and Importing Seating Data into LifeMOD™ Environment

There are three main steps in the Vicon System analysis. The first is patient preparation (Figure 1.15), where the anatomical measurements of the patient are taken and markers are attached according to standard marker sets. Second is the process of data recording (Figure 1.16).

Figure 1.14 Vicon motion capture laboratory.

Figure 1.15 Subject with markers attached.

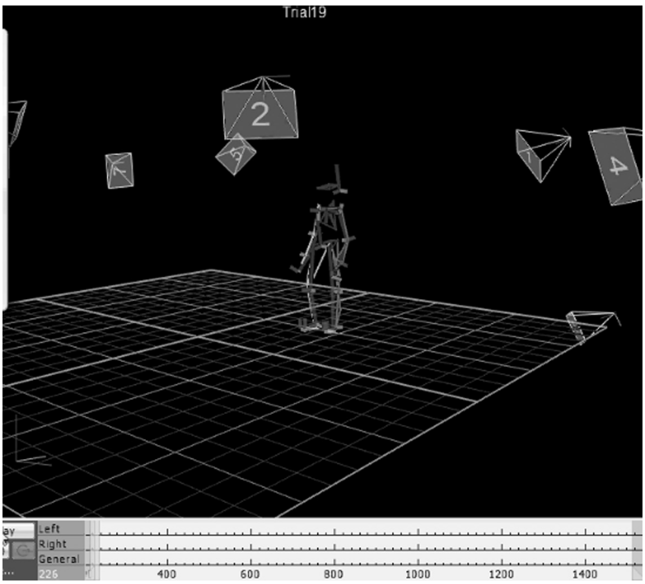

Figure 1.16 Recorded data system.

A static analysis is performed once, followed by the real motion. The motion is repeated many times until the best set of data is captured. Recorded data is then processed and imported into the LifeMOD™ environment (Figure 1.17). Vicon provides highly accurate movement information, which is an essential element of designing a realistic spine modelling system. Two subjects performed the trials. The Nexus MOCAP system was used to map the motion of the subject walking up to a chair, sitting down, and moving into a predetermined posture (leaning backward, straight, and leaning forward). The results were processed using Vicon Nexus and Vicon BodyBuilder. One set of results was used to try to achieve a workable model in LifeMOD™ using MOCAP import option. In this work, results from a model with discretised lumbar region are analysed.

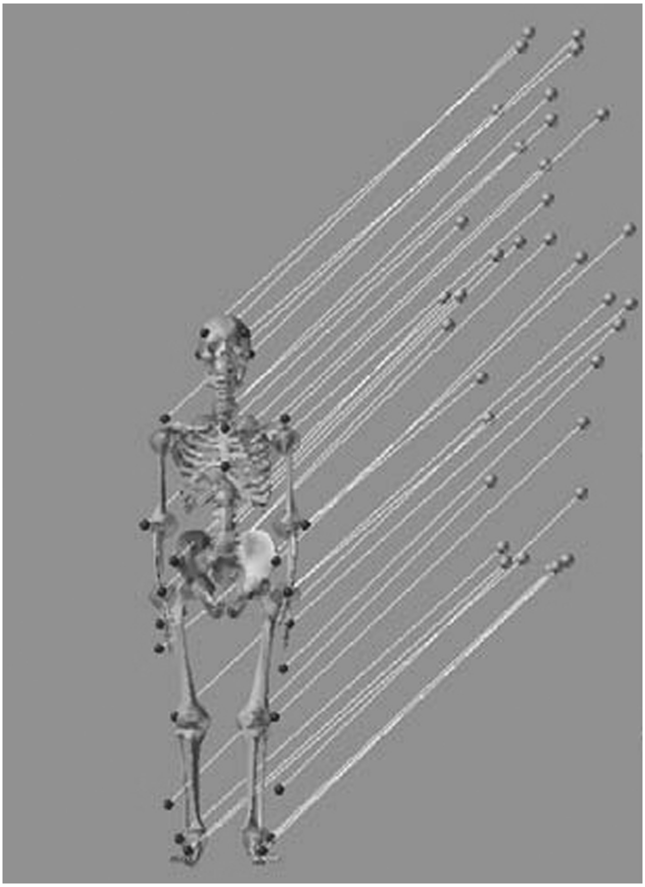

Figure 1.17 Import of Vicon data into LifeMOD™.

1.3.1.3 Effects of Different Postures on Spinal Forces

The subject is instructed to perform the following motion, as shown in Figure 1.18. Start at a normal gait (A) and approach the chair from the side; when the chair is reached turn (B) so that your back is to the chair (C); sit down (D), and then lean back to a normal sitting posture (E). Slowly lean forward and remain in that position for approximately 4–5 seconds (F). Afterwards, sit up straight (G), stand up (H), and walk forward.

For this study, the most important sections are D–H. The plot increases rapidly at D reaching a maximum of approximately 5850 N. The value drops to about 300 N when sitting reclined. As the subject bends forward, there is an increase to about 1200 N. As the subject leans back again, the force decreases to about 300 N again. When the subject stands up again, it reaches a peak of about 4350 N.

B, D, and H show the highest peaks. A (gait), C (standing), E, and G (both sitting in reclined) fluctuate around the same minimum force values in the plot. The values at sitting leaning forward show a very large increase in force (1200) from sitting in a reclined posture (300 N). This value remains almost constant over the duration of sitting in that particular posture and it is still lower than the maximum peak forces during D and H. Table 1.2 shows the peak values at various phases of motion. Note that the forces on the joint indicate the resultant force due to both external and internal forces that are interrelated and are established to allow the body to be mobile yet stable. The main external force is the ground reaction force (GRF, shown in Figure 1.19), which is the force exerted on the body by the ground and can be explained through Newton's Third Law. GRF is a reaction force to the forces of gravity (the weight of the subject as well as the internal forces generated by the muscles). The weight of the subject is approximately 580 N (59 kg × 9.81). The lumbar spine sustains the

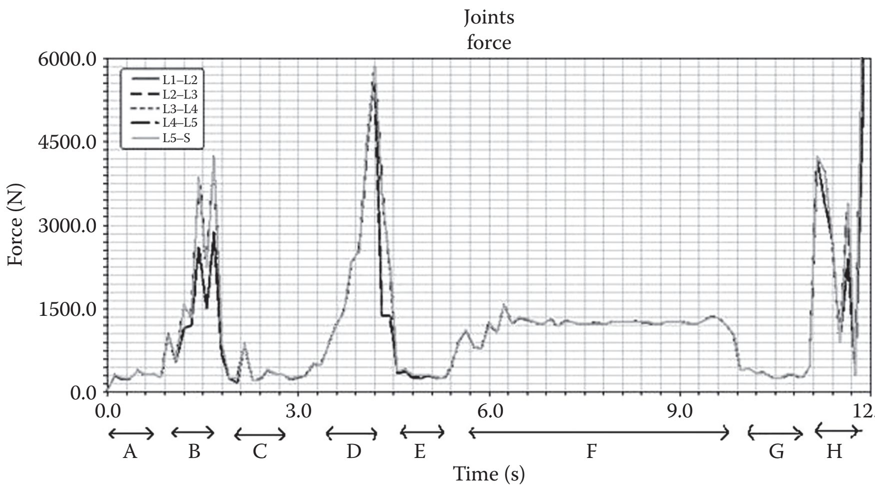

Figure 1.18 Force motion sequence.

Table 1.2 Summary of Joint Force Peak Value Data for MOCAP Analysis

Phase	D	E	F	G	H
Peak Value (*N*)	5850	00	1200	300	4350

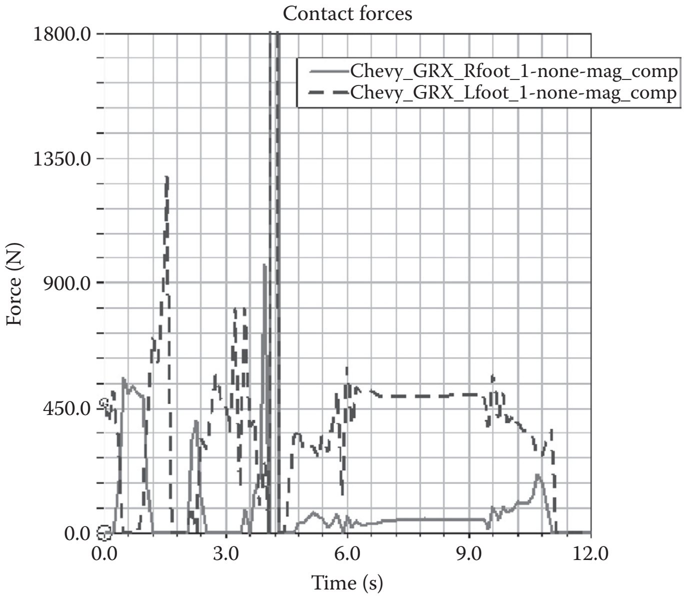

Figure 1.19 Ground reaction force plot.

loading of the upper body. This loading can change in tension, compression, torsion, or shear depending on the motion [22]. Overall, all five lumbar joint forces plotted follow the same trend, which indicates that the load is well distributed over the joints. Coupled motion (motion of one direction that affects the others) is also prominent as all respond the same way at the same time. This trend in Figure 1.18 can be explained as follows. In a normal gait (A), the body is upright and both the external and internal forces are stabilised to allow little fluctuation in the overall force.

During B, the body changes momentum both in direction and in magnitude. Increases in the joint forces indicate that the muscles actively respond to change in motion. Force is increased in torsion due to the twisting movement of the vertebrae relative to each other when the subject turns. At C, the subject is standing upright and erect with little movement. Forces are then stabilised and reach almost the same minimum as phase A. As the subject sits down, this bends his or her knees and actively contracts the leg muscles, which in turn exerts a higher force onto the ground. This increased force generation affects GRF and also the resultant joint force. The body also increases in flexion, which decreases lordosis and increases compression in the spinal joints. Forces rapidly decrease as the body adjusts to accommodate the change in loading. At E, the subject reclines, which increases lordosis once again and therefore decreases compression. As the subject leans forward with increased flexion, joint force increases as expected due to decreasing lordosis and increased compression. Also, shear loading is produced by translational movement of the vertebral bodies; as flexion occurs, vertebrae slide against each other [22]. As the spine bends forward, there is also an increase in the activity of the back muscles. If forward flexion increases, transition of the spinal load bearing from muscles to the ligamentous system takes place. Because of the downward direction of their action, as the back muscles contract, they exert a longitudinal compression of the lumbar vertebral column, and this compression raises the pressure in the lumbar intervertebral disc [23].

The constant force indicates that once a subject adjusts to a posture, there is little fluctuation in force. Therefore, in order to decrease the overall force, the subject must be supported externally. At G, the subject straightens up (force decrease) and stands up at H (increase caused by high GRF and muscles). In the plot for GRF versus time (Figure 1.19), the peaks correspond to the peaks in the joint-force plot. This indicates that high GRF is directly related to high force transmission to joints. It was observed that the subject's feet sometimes lost contact with the floor when sitting because the subject was short. At the first peak, the subject turns on the left foot, and therefore, all weight is supported on that foot until the right foot is placed beside it. At the second high peak, the left foot muscles are still regaining stability and possibly have higher muscle contraction (hence high GRF response). During the duration of seating, the left foot GRF is higher, perhaps because the posture is asymmetric, or because the subject favours one side or is forcing down on one foot to touch the floor. Perhaps putting in a footrest to properly support the body would distribute the force evenly and reduce the overall spinal loads.

The tensile forces in the erector spinae muscles as shown in Figure 1.20 correspond to the varying requirements of the muscle. The difference in left and right leg in seating particularly corresponds to a high GRF. There are numerous advantages in using MOCAP for these simulations. Firstly and most importantly, each stage in the motion can be clearly seen and defined in the graphs. Therefore, the force at the particular posture under study can be found quite easily (in this case, approximately 1200 N for leaning forward).

1.3.2 Studying and Comparing Biodynamic Behaviour of Spinal Fusion with Normal Spine Models

Spinal fusion has become a popular surgical procedure for chronic disabling back pain during the past 20 years but is widely considered to be a last resort as long-term complications can often arise due to the nature of the procedure. Although surgical procedures involving vertebral fusion produce a relatively good short-term clinical result in relieving pain, they alter the biomechanics of the spine. For example, they will immobilise the spine unit and reduce the spine's ROM. In addition, they can lead to further degeneration of the discs at adjacent levels.

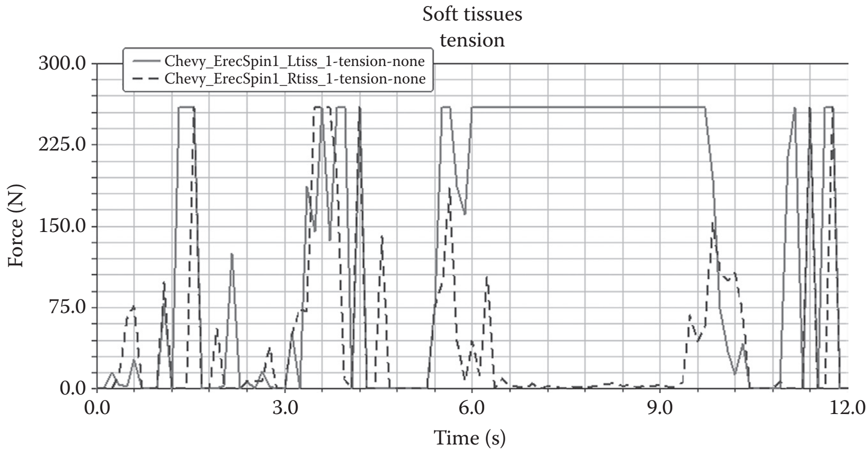

Figure 1.20 Left and right side muscle forces.

These problems can be verified by using the detailed spine model presented in Sections 1.2.2 through 1.2.6. In the present spine model, spinal fusion can be made at either the L3–L4 or L4–L5 level by applying fixed joints between vertebrae. In severely degenerated cases, these two levels are fused together. Then, external forces are imposed on a certain vertebra, and comparison between spinal fusion and a normal spine model can be achieved. Figures 1.21 through 1.23 show three cases of locomotion comparisons between the normal spine model and fusions at the L3–L4 level, L3–L4 and L4–L5 levels, and L3–L4 and L3–L4–L5 levels, respectively.

1.3.3 Modelling of Spine Deformity

An arbitrary three-dimensional spinal deformity can be described by a combination of the deformities in three spatial planes, that is, the frontal (coronal), sagittal (lateral), and transverse (axial) plane. Each deformity can be characterised by the corresponding spinal curvature and vertebral rotation [24]. Based on these characterisations, three deferent types of spine deformity can be defined.

1.3.3.1 Kyphosis

Kyphosis is an exaggerated backward spinal curvature in the sagittal plane, characterised by a humpback appearance (see Figure 1.24).

1.3.3.2 Scoliosis

Scoliosis is a medical condition in which a person's spine is curved from side to side in the coronal plane. Although it is a complex three-dimensional deformity, on an

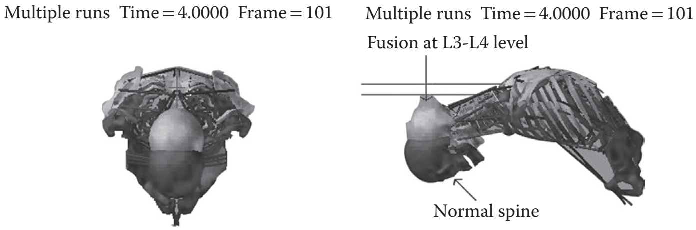

Figure 1.21 Normal spine model compared with L3–L4 fusion.

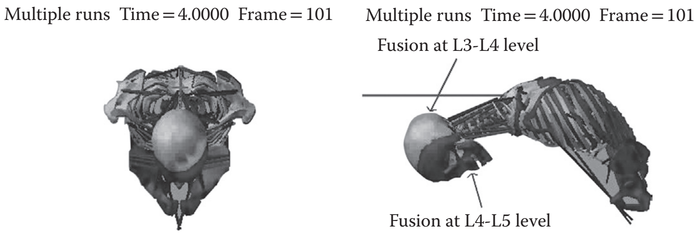

Figure 1.22 Fusion at L3–L4 compared with L4–L5 fusion.

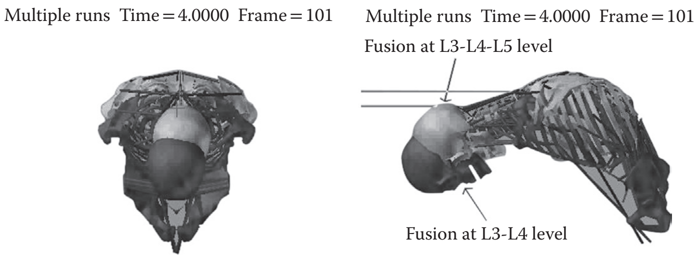

Figure 1.23 Fusion at L3–L4 compared with L3–L4–L5 fusion.

X-ray, viewed from the rear, the spine of a person with scoliosis may look more like an 'S' or a 'C' than a straight line [25].

1.3.3.3 Kyphoscoliosis

Kyphoscoliosis is an abnormal curvature of the spine in both the coronal and sagittal planes. It is a combination of kyphosis and scoliosis [26].