APPLICATION OF SPORTS BIOMECHANICS FOR LOWER LIMB AMPUTEES

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Introduction

Sports participation and prosthetic design for physically handicapped people are largely influenced by biomechanics research. In fact, the knowledge of mechanical physics and biological material properties gave birth to the sciences of biomechanics and kinesiology. Through these sciences, significant contributions have been made in understanding basic human movements, the goal-oriented movements as well as the adapted human movements (Kreighbaum and Barthels, 1985). Biomechanics research has provided data leading to advancements in equipment and prosthetic design (Becker, 1984; and Burgess, Hittenberger, Forsgren and Lindh, 1983).

In the United States there are 1.7 amputees per 1000 persons (Glattly, 1964) and 59% of the lower limb amputations are performed at the below the knee level (Kay and Newman, 1974). The 1980's were marked by a national concern for physical fitness and this need is even more important for the physically handicapped (Kegel, Carpenter and Burgess, 1978; Martel and Estok, 1984). Most of the research dealing with amputees has been focused on daily living activities (Isakov and Becker, 1985; Wagner, Sienko, Supan, and Barth, 1987) and traditional prosthetic feet were initially designed only for walking (Campbell and Childs, 1980; Inman, Ralston, and Todd, 1981). Kegel et al. (1978) reported that 61.2% of lower limb amputees were active in at least one physical activity; their research showed that the most popular avocational activities were fishing and swimming while the most difficult were running and long distance walking. Miller (1981-1984) performed extensive studies on the development and improvement of running skills in below the knee amputees. At the same time, more functional prosthetics such as ultra-light feet (Wilson and Stills, 1976), water safe feet (Koester, 1983) and rock climbing feet (Levesque and Gauthier-Gagnon, 1987) were developed in order to meet the growing and more specific needs of sport oriented amputees. Research on sports prosthesis is progressing mostly in the areas of knee restoration (Zarrugh and Radcliffe, 1976) and the ankle functions as in push-off. Different systems and materials are used to imitate the activity of the gastrocnemius and soleus muscles in walking or running. These prostheses are commonly known as Energy Storing Feet (Michael, 1987) but can be properly termed as the Dynamic Elastic Response (D.E.R.) system which includes the Seattle foot (Burgess et al, 1983; Hittenberger, 1986; and Reswick, 1986), the Dual Ankle System (D.A.S.) foot (Voisin, 1987), the Flex foot (Bach and Wooley, 1986; and Wagner et. al, 1987), the Terry Fox Prosthesis (Martel and Estok, 1984) and the SAFE foot (Campbell and Childs, 1980). Clinical comparisons have already been made concerning the efficiency of these foot prostheses for sports participation (Michael, 1987). However, very little objective information has been reported in amputees research on the shock components of the foot impact, even though some authors have reported quantitative analysis of the longitudinal foot-ground impact in normal populations. Physical activities in which repetitive shocks are transmitted to the rest of the body are considered risky (Prince, 1982; and Therrien et al, 1982) since they can affect the biological integrity of the body and can lead to degeneration of the cartilage (Finlay and Repo, 1979) as well as to stress fractures and osteoarthritis (Light et al 1980).

Furthermore, since the intensity of impact increases with the displacement speed (Stein, Charles and James, 1988), Light et al. (1980) suggested that there is no body readjustment possible to absorb impact transmission due to the very short duration of the phenomenon.

Case Study

The present case study was performed on two subjects; the first was a female below the knee amputee aged 19 years who was amputed at age 3 years. The second subject was a male above the knee amputee aged 23 years who was amputed 9 months before the study. In both cases, amputation resulted from a traumatic condition. At the time of the study both subjects were healthy and played regular or adapted sports.

Method

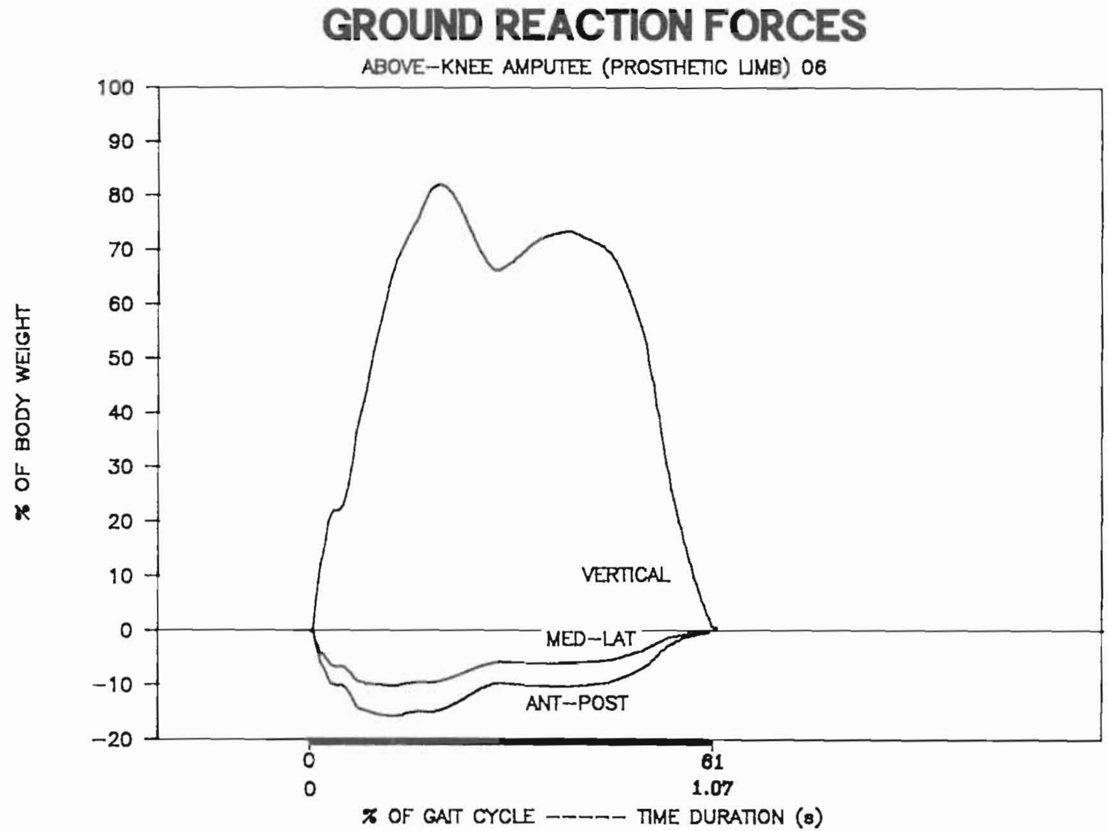
The subjects were to walk at normal speed on a 10-meter walkway and were to contact a force platform with the prosthetic limb (PL) and the sound limb (SL). Kinetic data were obtained through a tridimensional force platform and two tri-axial accelerometers placed at the ankle and hip. In order to obtain kinematic data, self-adhesive markers were placed on the subjects, in order to estimate the joint centers. Ten sequences for each leg were taken with two video cameras. The first camera (set at a variable shutter speed of 1/250 s) was placed perpendicular to the movement plane. The second video camera was placed 8 meters in front of the force platform. The vertical acceleration data and the ground reaction forces were collected and computed with an IBM-PC computer. The kinetic and kinematic data are presented in Table 1.

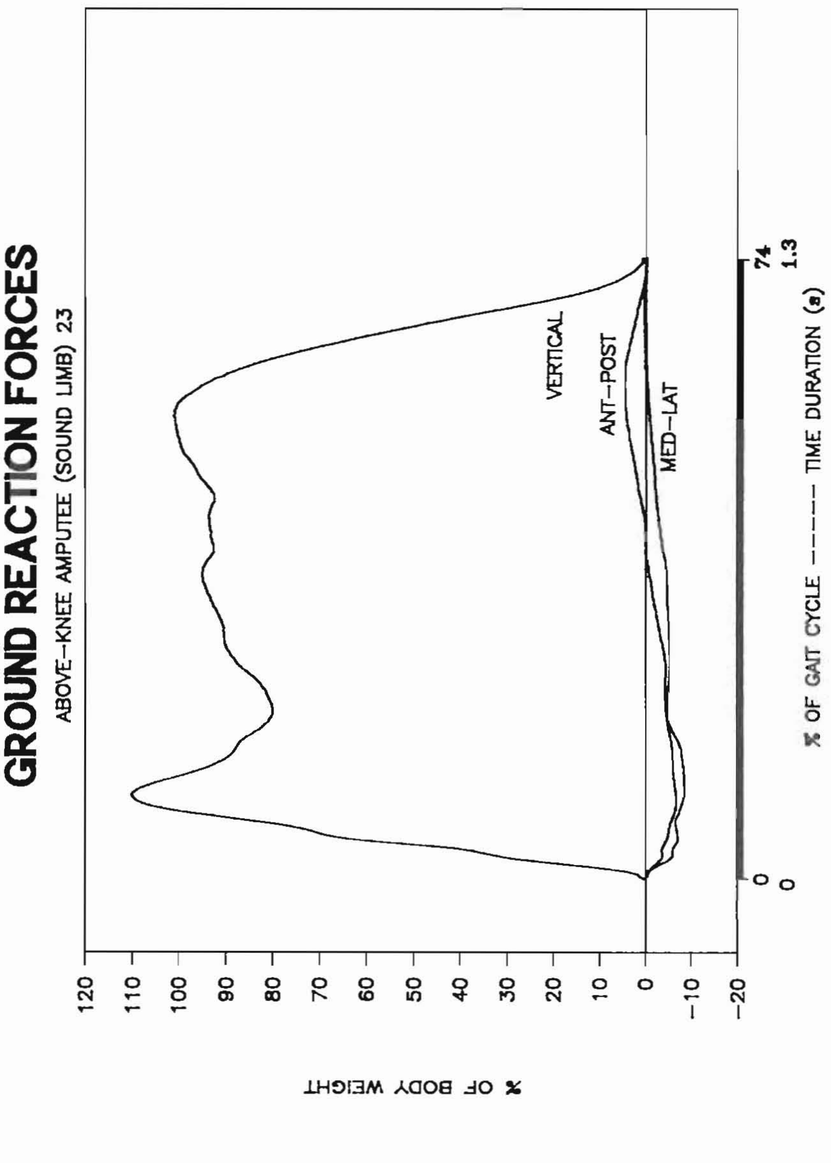
Kinetic and Kinematic Data on Walking Gait in
Above and Below-Knee Amputees

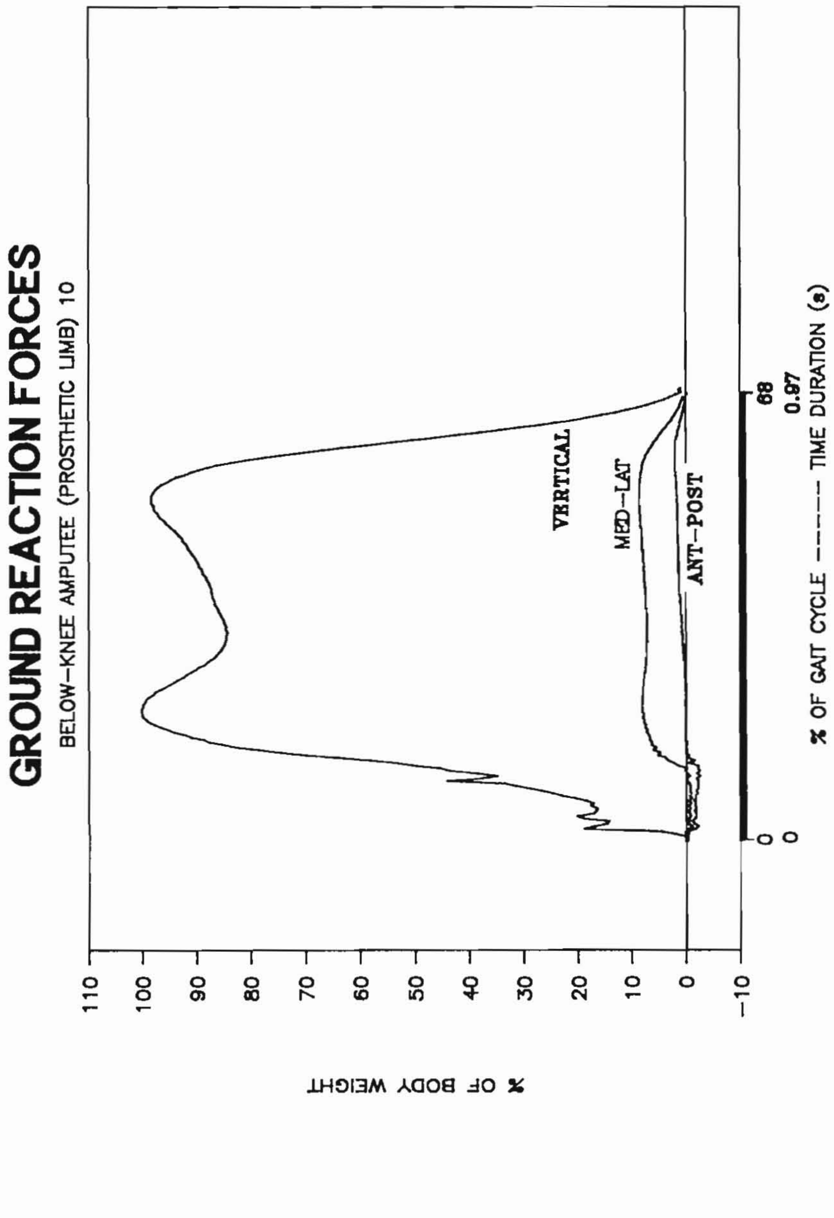
Parameters	Conditions	
	A-K	B-K
Kinematic Data Walking speed	0.62 m/s	0.72 m/s
Gait Cycle		
Total Time	1.75 s	1.43 s
Swing Phase		
Sound limb	0.46 s	0.44 s
Prosthetic limb	0.69 s	0.45 s
Stance Phase		
Sound Limb	1.30 s	0.98 s
Prosthetic Limb	1.07 s	0.97 s
Double Support	0.27 s	0.29 s
Trunk Deviation (degrees)	8	4
Angels (degrees)		
At Heel Contact		
Prosthetic Limb	8	28
At toe Off		
Prosthetic Limb	24.5	48

The main difference between the two subjects was the motion of the body during the swing phase of the prosthetic limb. The loss of the knee joint affects the dynamics of the entire locomotion system. Another important difference was the amount of trunk deviation measured during the swing phase of the PL. The strides were of the same length during gait cycle for both subjects during the total gait cycle but were different for swing phase and stance phase for the AK amputee. The angle between the ground and the foot were different between BK and the AK conditions during heel contact and toe off: 28 and 48 degrees, respectively for BK and 8 and 24.5 degrees for the AK.

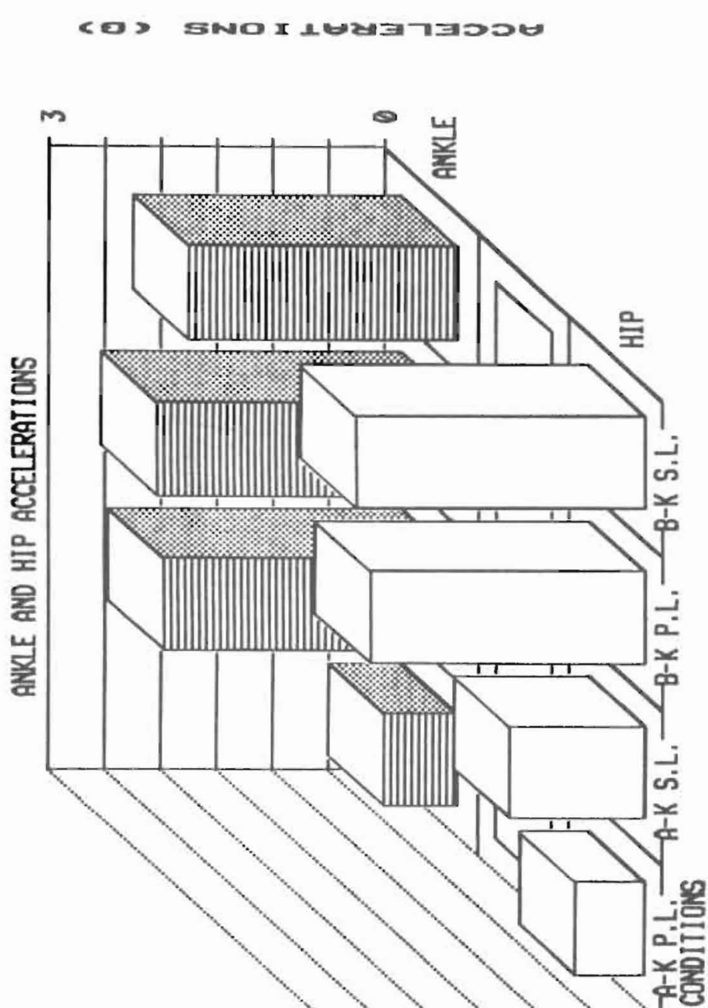
The vertical ground reaction forces recorded showed that the AK and BK subjects load respectively 82% and 100% of the body weight on their PL, but that they applied 110% and 150% of the body weight on the SL. Further, there is a difference in the medio-lateral force component in that the AK amputee applied more load toward the inside while the BK had a higher load toward the outside. When analyzing anterior-posterior force, the AK amputee displayed only a braking component while the BK amputee had a little push off at the end of the stance phase. Vertical accelerations both at the ankle and at the hip were quite low due to the slow walking speed.


GROUND REACTION FORCES
ABOVE-KNEE AMPUTEE (PROSTHETIC LIMB) 06
% OF BODY WEIGHT
100
90
80
70
60
50
40
30
20
10
0
-10
-20
VERTICAL
MED-LAT
ANT-POST
0
61
0
1.07
% OF GAIT CYCLE ----- TIME DURATION (s)


GROUND REACTION FORCES
ABOVE-KNEE AMPUTEE (SOUND LIMB) 23
VERTICAL
ANT-POST
MED-LAT
% OF BODY WEIGHT
120
110
100
90
80
70
60
50
40
30
20
10
0
-10
-20
0
0
74
1.3
% OF GAIT CYCLE ------ TIME DURATION (s)



GROUND REACTION FORCES
BELOW-KNEE AMPUTEE (PROSTHETIC LIMB) 10
VERTICAL
MED-LAT
ANT-POST
110
100
90
80
70
60
50
40
30
20
10
0
−10
% OF BODY WEIGHT
0
0
68
0.97
% OF GAIT CYCLE ------- TIME DURATION (s)


ANKLE AND HIP ACCELERATIONS
ACCELERATIONS (G)
3
0
ANKLE
HIP
A-K P.L.
A-K S.L.
B-K P.L.
B-K S.L.
CONDITIONS

Discussion

This preliminary study consisted of an analysis of two amputee conditions during walking at normal speed. The results showed not only major differences in the gait pattern of the lower limb but also in the x,y,z ground reaction forces, as well as in the vertical accelerations at the ankle and at the hip.

The time lapse between amputation and evaluation was a factor that was considered in the performance: the BK amputee did not require any assistance and was able to practice many sports like volleyball, badminton and running. Contrarily, the AK subject, a recent amputee, needed a cane to assure himself from a fall. The loss of one or two articulations replaced with a prosthesis has a theoretical influence on the quality of shock absorption due to the reduction in the biological tissues (Miller, 1981) as well as the loss of joint functions (Miller, 1981). The intensity of shock transmission through the prosthetic limb is more severe during sport locomotion activities (Miller, 1981). The reduction of the intensity of shock transmitted is essential to the protection of the person's muscular skeletal system. Furthermore, the absorption system has to be used without affecting locomotion stability.

Conclusion

Prosthetic equipment design is a combination of materials and engineering knowledge. In the future, further research will be carried out on the multi-axis function of the ankle including eversion and inversion of the foot (Campbell an d Childs, 1980; and Voisin, 1987).

Our evaluation of the BK and AK amputees, being solely based on two subjects, is insufficient to allow generalization to the overall amputee population. These results do however lead us to a better understanding of the locomotion patterns and to the main differences between the two conditions. Physical therapists, researchers and prosthesists must realize the possibility that active and healthy amputees are able to run and practice sports (Burgess et al, 1985; Kegel et al, 1978; and Martel and Estok, 1984). However, limitations still lie in the integrity of the residual limb and in the foot-ankle system. Finally, extensive research in shock transmission of sports oriented amputees must be conducted in order to prevent injuries.

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