ANALYSIS OF THE PERFORMANCE OF ABOVE-KNEE AMPUTEES IN CLIMBING STAIRS

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INTRODUCTION

Although the efforts to improve the movement capability of amputees date back a long way we are still quite unable to guarantee their full rehabilitation and there is plenty of room to improve the means for evaluating the efficiency and comfort with which the prostheses are used.

The analysis of the amputee's gait has been done according to the following main approaches: (a) Time-distance parameters of gait; (b) Motion analysis; (c) Force plate analysis; (d) Energy factors in gait. All these approaches have been used for both below knee amputees (BKA) and above knee amputees (AKA) and their performances compared with those of able bodied (AB) subjects.

The first addresses the quantification of gait parameters such as the gait speed and rhythm, the stride length and duration and the stance phase duration.

The second approaches stems from the bhelive that a better understanding of the complete phenomenon of locomotion in amputees could be attained by performing motion analysis studies which, besides the legs, include other body segments. Engsberg and collaborators (1992) compared the position of the center of mass (CM) in AB and BKA children finding that, as a result of a forward bending of the trunk in the sagittal plane, the CM is lower and further forward for the BKA children.

The use of force platforms to measure the ground reactions are exemplified by the work of Suzuki (1972) who studied AKA and BKA patients and found the floor reaction for the artificial and normal limbs to be different.

The energetic cost of locomotion for BKA walking at normal speed has been found by different authors to be about three times larger than that for AB subjects.

The studies of the kinematics and dynamics of the task of climbing stairs are still very limited (Shinno, 1971; Andriacchi et al., 1980; McFadyen & Winter, 1988) and none was found dealing with AK subjects. In the present study a group of AKA is compared, using the time-distance parameters of the gait approach, with another of AB individuals in the performance of two locomotion tasks, namely, walking on level ground and climbing a staircase.

METHODS

Two groups, one of AB and another of AKA, of three individuals each, were video recorded performing the tasks of level ground walking (LG) and of climbing a stair case (SC). The video recordings were performed at 25 frames per second with two cameras located at a right angle to film the frontal and the sagittal planes, respectively. Reflective markers were placed on the spine at C7 and L5 levels and at the segment endpoints. The three dimensional co-ordinates of these marks were obtained with the help of a numerical package developed by ourselves. The synchronisation of the cameras was obtained by filming simultaneously a digital clock with a definition of a hundredth of a second. The CM is located by using a cylindrical segment model of the human body. The position of the CM is referred to a moving system of three orthogonal axes defined as follows: the X-axis passes through the markers placed over the sides of the pelvis and is horizontal and parallel to the floor surface, the Y-axis is also horizontal and is perpendicular to the X-axis passing through the mid point between the two markers and the Z-axis is the vector product of the two previous axes. All the distances and co-ordinates are normalised by deviding them by the subject's height.

RESULTS AND DISCUSSION

The results presented in this paper refer to a representative task length which is a complete step cycle, in the first case, and the climbing from the first to the third step in the second case. The data analysis is performed by digitising sequences of twenty gait strides for each subject and for the two tasks, which they all performed, and by computing the corresponding ensemble averages.

The analysis of the results shows clearly that the CM of the AKA tends to be located at a lower level and lagging slightly behind along the Y-axis relative to that of the AB subjects and, also, that it tends to be always located on the prosthesis side of the X-axis

Figure 1 illustrates the evolution of the forward (Ω) and lateral (θ) bending angles throughout the LG stride cycle. These diagrams show that the values of Ω tend to oscillate around 4 degrees in the case of the AKA, and that the AB subjects show a deeper variation in the amplitude of this angle thus revealing a greater postural equilibrium control. The values of θ are, as it is to be expected, notoriously larger in the case of the AKA subjects reflecting the fact that these subjects do make use of this oscillatory movement to guarantee the firm support on the prosthesis to propel themselves through the stride cycle.

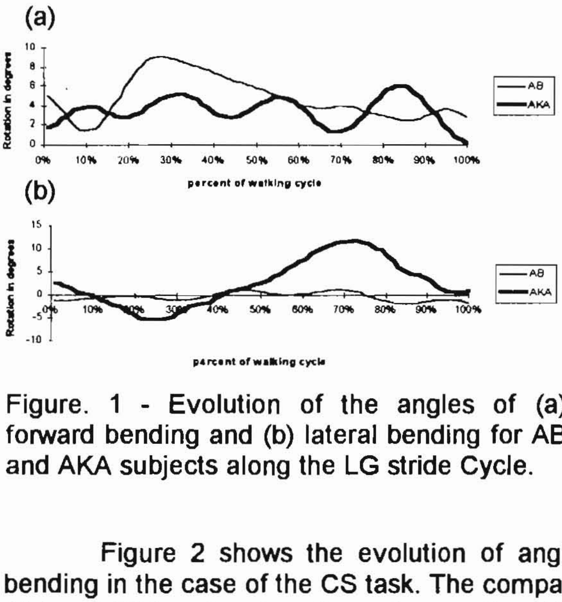

Figure. 1 - Evolution of the angles of (a) forward bending and (b) lateral bending for AB and AKA subjects along the LG stride Cycle.

Figure 2 shows the evolution of angles of forward (Ω) and lateral (θ) bending in the case of the CS task. The comparison between the diagrams of figs. 2(a) and 2(b) make clear that, whereas the AB subjects tend to reduce drastically the amplitude of the sagittal plane oscillations, the AKA increase substantially that oscillation with the obvious purpose of facilitating the climbing of the stairs. Insofar as the oscillation in the frontal plane is concerned, both groups of

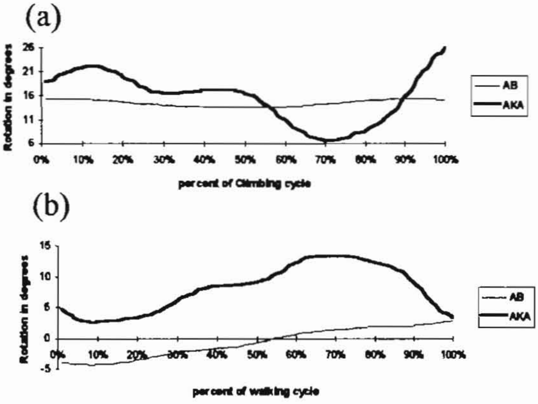

Figure. 2 - Evolution of the angles of (a) forward bending and (b) lateral bending for AB and AKA subjects along the CS cycle.

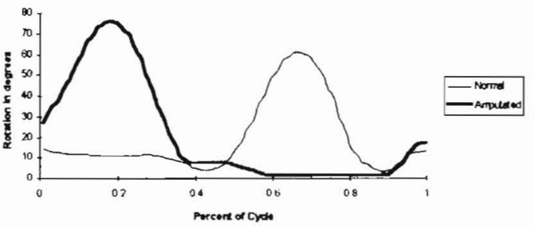

Figure. 3 - Evolution of the angle of flexion-extension for AKA subjects performing LG walking.

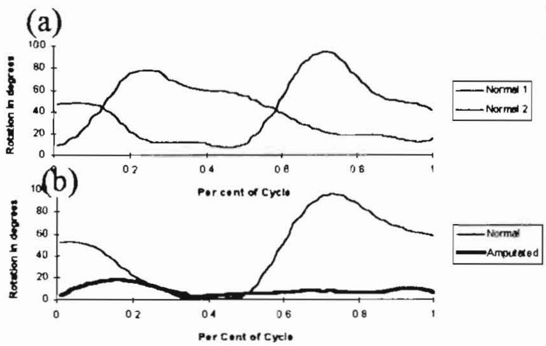

Figure. 4 - Evolution of the angle of flexion-extension for AKA subjects performing a CS task (a) AB subjects (b) AKA subjects.

participants show an increased value of its amplitude but differ in the fact that in the case of the AKA the oscillation occurs around a bias inclination of approximately 8 degrees to the side of the prosthesis.

Figure 3 illustrates the evolution along the stride cycle of the sagittal knee flexion-extension angle of both legs of AKA subjects for LG walking. The cycle of the normal leg of the AKA differs little from that of both legs of AB subjects. The normal leg begins the cycle with a larger flexion angle value and sustains it during most of the first phase of flexion-extension which corresponds to the stance phase; this behaviour reflects the practice of shorter and more conservative stride length. The second phase of the cycle for the normal leg is similar to that of the AB legs but starts slightly later and lasts less. The amputated leg, on the other hand, has practically no flexion during its stance phase and the phase of maximum flexion is wider and reaches a peak value of approximately 75°, a value which not only exceeds that reached by the healthy leg but also is bigger than those by both legs of AB subjects.

Figure 4(a) illustrates the stride cycle for AB individuals performing a CS task and, although both legs display two flexion-extension phases, this phases differ appreciably suggesting that the leg which reaches smaller maximum flexion values but has a longer flexed phase is responsible for a larger measure of the climbing impulsion.

Figure 4(b) shows clearly that the AKA group is only able to get climbing impulsion from the normal leg which has a cycle similar to those of the AB group differing only in the values of the minimum flexion angle.

CONCLUSIONS

This study clarifies certain important differences between the AB and AKA locomotion patterns. In the first place, it shows that the amputee tends to indulge in larger trunk oscillations, both in the sagittal and in the frontal planes, oscillations which correlate closely with the more demanding equilibrium and propulsive needs. This behaviour involves naturally a greater energy expenditure and that more attention is dedicated to the performance of the task. Furthermore, the trunk movements in the sagittal plane are accompanied by head rotation movements in the same plane which are needed to allow the frequent gaze at the ground ahead in order to compensate for the lack of sensorial influx from the missing limb.

The analysis of the exercise of climbing stairs has shown performance differences which are much wider than they are for level ground progression, thus reflecting the fact that it corresponds to a more demanding mechanical task. However, walking up and down ramps, stairs and other architectural barriers is a common fact of life for everyone including amputees who, therefore should be trained to use their prosthesis adequately in all situations they have to face normally. The evaluation of that adequacy is in itself a quite difficult task and we can only hope to reach meaningful results when the amputee is tested in a large range of locomotion situations. The most relevant conclusion of the present work refers precisely to this need of complex training and evaluation regarding the proper use of prostheses by amputated individuals.

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