STRATEGIES FOR WALKING ON LOW-FRICTION SURFACES

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INTRODUCTION

Those abnormal gait patterns which are caused by either a predominantly efferent or a predominantly afferent sensorimotor deficit show common features - a reduced step length and an increase in both the foot-flat duration and in the double-support to single-support ratio[1]. Similar changes occur in healthy subjects when there is a real or perceived threat to stability, for example during beam walking[2]. The changes observed in both studies indicate that subjects adopt a protective strategy which maximizes the period of double-support and foot-flat contact, and reduces the velocity of gait. As the difficulty or complexity of the locomotor task becomes greater, the threshold at which subjects adopt the protective strategy is lowered'. The common features of the protective strategy and the fact that it is executed automatically suggest that it is a response mediated by the central nervous system in order to both stabilize balance by adjusting the gait pattern and to reduce the potentially destabilizing effects of proprioceptive reflex responses. These changes are made at the expense of gait velocity.

Under conditions where walking subjects encounter a real or perceived risk of instability, for example whilst traversing a slippery surface, the precise control of body movements and forces is likely to be especially critical. We predicted that this would result in healthy subjects adopting similar and consistent patterns of movement and force control. Of particular interest are the initial-contact and push-off phases since these are the periods where the risk of slipping is greatest'. This paper details the development of the APRE low-friction walkway and summarizes the results of initial experiments to investigate force and movement patterns of subjects during normal and low-friction gait. The results obtained may assist in the development of specialist footwear designed to maximize gait performance whist maintaining protection against the hazards associated with slips and falls.

METHODS

A walkway 10 m long and 2.5 m wide was established on a linoleum-covered floor. Mounted flush with the surface, midway along the length, and offset to one side of the central line of the walkway, is a multi-component force platform (Kistler 9821B12). The platform is connected to its charge amplifier (Kistler 9865A) by a buried cable and is controlled by a laboratory interface (Cambridge Electronic Design 1401) connected to a personal computer. Countersunk bolts and screws were used to locate 3 mm thick poly-tetra-flouroethane (PTFE) sheet, measuring 0.8 m by 3.24 m, to the central area of the walkway. To ensure force. isolation between the platform and the surrounding floor surface, a 2 mm wide gap was established in the PTFE sheet around the circumference of the force platform. Finally, the PTFE sheet was cleaned using household fumiture polish. A load-cell (Entran ELF-1000-100) was used to quantify the error due to deformation of the PTFE sheet.

The gait of six healthy subjects was studied under three conditions: during normal gait over a linoleum floor; during the transition from the linoleum floor surface to the PTFE-surfaced force platform; and during established gait on the PTFE-surfaced walkway. The coefficient of friction (μ) between the floor surface and the sole of the foot was 0.7 on the linoleum; $\mu=0.8$ on the force platform and $\mu=0.3$ on the PTFE sheet. In each condition at least 15 steps were recorded from periods of established gait. Subjects were dressed in casual clothing and wore only thin polyester/wool mix socks. Retro-reflective markers (diameter 37 mm or 22 mm depending on location) were fitted to a maximum of IS anatomical landmarks. Subjects were illuminated by floodlights as they traversed the walkway at their own speed. Video-tape records of movement were made using three frame-synchronized cameras. Force data were collected on a separate recorder (Biologic 2816 16-channel DAT). The camera synchronization (genlock) pulses and an event trigger (a push-hutton switch which produced an LED flash in each camera view together with a concomitant positive-going 50 ms square wave pulse) were also recorded on the DAT to assist subsequent synchronization of movement and force data.

Analysis of the video taped records was performed using a Peak Performance Technologies video analysis system. To permit frame identification. time-code was dubbed onto the audio track of each pre-recorded video tape using a video recorder (Panasonic AG-7330) under computer control. The moment of initial-contact of each step of interest was identified and all the individual video frames (numbering between 50 and 70 depending on stride duration) corresponding to each of the steps were digitized. The x, y and z coordinates of each anatomical marker were calculated and stored by the computer. Each channel of force data was digitized at 500 Hz and the sum of the vertical (F_v), anterior-posterior (F_{a-p}) and medial-lateral (F_{m-l}) force components was calculated at each sample interval. The data were presented graphically as force vectors and time-histories together with the frame number of the corresponding video record of movement. Waveform averaging was performed using a **Lotus123** spreadsheet.

RESULTS

Consistent differences were observed between the force profiles of normal gait and those obtained during low-friction gait. Under normal conditions F_v showed the typical profile of two peaks (associated with the weight acceptance and push-off phases). $F_{a\text{-}p}$ was negative during weight-acceptance and positive during push-off. During low-friction gait F_v was reduced and there was not a well-defined trough in the force profile during mid-stance due to F_v being maintained at high levels throughout the stance phase. The amplitude of the $F_{a\text{-}p}$ negativity following initial-contact was reduced during low-friction gait as was the peak-to-trough amplitude of F_v ($P < 0.01$). These changes in force profile are similar to those observed during normal gait when there is a reduction in gait velocity[4] but the magnitude of the changes measured in this study are too great to be accounted for by velocity differences alone. In addition, the profile of F_v during mid-stance is quite different from that which occurs during normal gait, even at very slow walking speeds.

Immediately before initial-contact between the foot and the walkway surface the knee was held in a slightly flexed position. On weight-acceptance the knee briefly flexed (knee-yield) before extending. The ankle was held in a dorsiflexed attitude during late swing phase and underwent a controlled plantarflexion during early stance phase which resulted in the entire sole of the foot being brought into contact with the ground more rapidly than is normal. Under low-friction conditions knee flexion was increased prior to weight-acceptance, and ankle plantarflexion was greater. Following initial contact, the amplitude of knee-yield was reduced ($P < 0.01$). Throughout the step-cycle on the low-friction surface the trunk was held flexed on the thigh by up to 15° more than during normal gait with the result that the centre of mass of the body was moved forward.

CONCLUSIONS

Consistent and statistically significant changes have been observed in body posture and the force time-histories during low-friction gait when compared to gait on a normal surface. These changes are in broad agreement with the temporal changes previously described[1,2] and support the notion that a protective gait strategy is adopted when stability is threatened. The changes in gait which result from this protective strategy affect the whole-body and involve adjustments not only to temporal factors but also to body kinematics and the development of ground reaction forces. The protective strategy minimises the risk of slipping when sudden limb loading occurs during early stance phase. The combined effect of the force and postural changes observed during early stance is to reduce the vertical acceleration and the forward velocity of the body. In summary, the protective gait strategy adopted under low-friction conditions is a whole-body response which maximizes stability at the expense of gait velocity.

Slippery surfaces are usually encountered where a smooth floor surface is contaminated with liquids or greases. Hospitals, swimming baths, meat factories, garages etc. all have an established risk of slipping. The biomechanical study of the response to slips, and to the threat of slips may elucidate the mechanism which determine whether an individual will slip on a given surface. To minimize the risk of slipping the gait pattern can be altered to maximize stability or materials employed to increase friction between the walker and the floor. Such materials may take the form of anti-slip floor surfaces or be a component of footwear. Techniques similar to those described here will enable the effectiveness of these or alternative measures to be quantified..

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