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„Artificial Feedback for Transfemoral Amputees - Theoretical Analysis“

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Artificial Feedback for Transfemoral Amputees - Theoretical Analysis

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Introduction

More than half of all lower-limb amputees suffer from falls in the first year after receiving their prosthesis [1] and have a lower balance confidence in general [2]. In part, this is due to actuation constraints of current prostheses. However, loss of a limb also results in impaired sensory function. While the sensory-motor loop of able-bodied humans can constantly rely on cutaneous feedback from the foot sole as well as proprioceptive information from muscles and joints, this feedback is missing in amputees. They are only provided with sensory information from their stump (Figure 1). It is hypothesized that this limitation is in part responsible for balance problems of amputees [3].

A promising approach to provide amputees with similar information is to use artificial sensors in the prosthesis and to display these signals to the amputee, for example using auditory, visual or tactile displays. This method is also called sensory substitution. Several studies with lower-limb amputees have been reported in the literature; while the limited number of users allows no conclusive statements, the center of pressure (CoP) at the foot sole was reported to be a useful signal [4, 5], as could be expected from studies with able-bodied subjects that investigated the role of the CoP in balance control [6]. Other researchers have found remarkable sensing capabilities in the stump and hip of transfemoral amputees, e.g. amputees were able to "feel" when their prosthetic knee joint was moved as little as 3° [7]. However, no quantitative analysis has been reported to assess whether additional sensory feedback could be useful for balance during standing or gait.

In this paper, we analyze how accurately a transfemoral amputee can determine the CoP based on forces in the stump. For that purpose, we developed a simple model of a transfemoral prosthesis and amputee perception based on the literature. Our goal is to compare the modeled perception of amputees to values from able-bodied subjects.

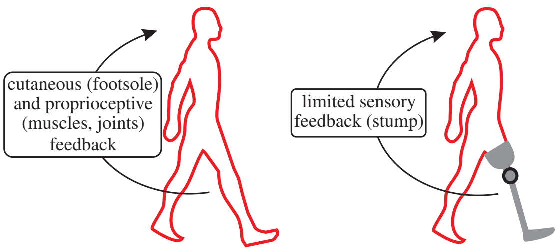

Figure 1: Limited sensory feedback of an amputee (right) compared to an able-bodied human (left).

Materials and Methods

We modeled the stance phase of the prosthetic leg as follows: We considered the three segments foot, shank and socket of a prosthesis. Since the knee joint remains fully extended during almost the whole stance phase for conventional prostheses [8], we considered the socket and shank to be a single segment (of length l). Furthermore, we modeled the ankle joint as a rigid connection at 90° to the leg angle, as is largely the case in amputee gait [8], such that the whole prosthesis is one rigid body. We also assumed that inertial forces acting on the prosthesis are negligible, since they are much smaller than external forces acting under the foot, and we assumed the prosthesis mass to be zero; this reduces the analysis to the static case, very similar to the balance task during quiet stance. If we assume the amputee can perfectly perceive forces $F_{S,x}, F_{S,y}$ and the torque τ_S in the stump (Figure 2), the center of pressure can theoretically exactly be determined:

$$\mathrm{CoP} = \frac{\tau_S - lF_{S,x}}{F_{S,y}}$$

However, several uncertainties prevent amputees from such accurate estimation of the CoP. All these uncertainties were modeled as zero-mean Gaussian noise with standard deviations (SD) derived from the literature: Proprioception of the hip angle is not perfect (at least $\pm 0.5°$ uncertainty [7]); in our model this is represented by a deviation from the simplified case where the prosthetic leg is completely vertical (θ in Figure 2). It has been shown using ultrasound that during stance phase, the residual femur moves significantly inside the prosthetic socket (e.g. $\pm 6°$ [9], ϕ in Figure 2); however, this depends on many parameters such as amputation level, muscle strength and socket fit. It is also unknown to what extent this movement is perceived by the amputee. We, therefore, modeled a lumped uncertainty in hip angle perception as noise with a SD of 2° that was used to transform ground reaction forces ($F_{gr,h}$, $F_{gr,v}$) to the prosthesis coordinate frame. Perfect moment and force sensing in the stump is also an oversimplification; the minimum threshold for a perceived force change was assumed to be 5 % of the applied force; due to lack of data for amputee stump perception, we made this assumption based on studies on weight discrimination in healthy subjects, which was at best 5 %, and often higher [10]. This was modeled as noise with a SD corresponding to 5 % of $F_{S,x}$, $F_{S,y}$ and τ_S.

We used data from a standard gait analysis (optical tracking system, ground reaction forces) of an able-bodied subject as reference values for the ground reaction forces and

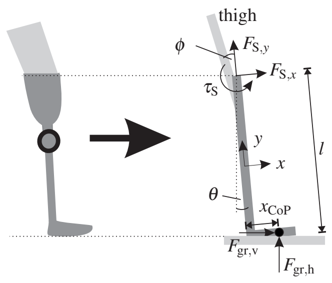

Figure 2: Simplified prosthesis model in stance phase.

the center of pressure. We used these values to calculate resulting stump forces and moments in our model. Then, we added the noise levels described above and conducted a Monte Carlo analysis using 500 simulations with different sets of noise. The SD over the resulting CoP trajectories was calculated as an indicator for a perception uncertainty of the CoP. This uncertainty was added to the actual CoP trajectory to put the values into context.

Results

Over the whole stance phase, the uncertainty in CoP perception remained fairly constant (Figure 3); it was slightly lower in the beginning and at the end of the stance phase, where the ground reaction forces are slightly smaller. The average SD of the CoP over the whole stance phase was 27 mm. This means that based on our model and the corresponding assumptions, amputees would be able to estimate their current CoP with about that accuracy.

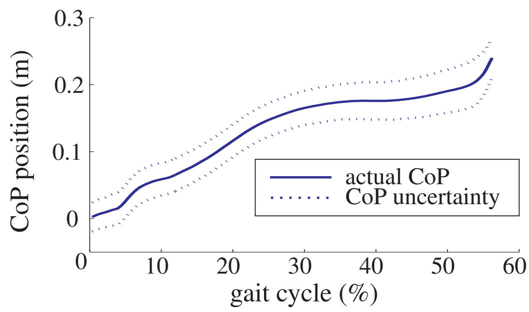

Figure 3: Stance phase uncertainty in CoP perception.

Discussion

The sensing of the CoP was substantially worse (27 mm SD) than the two-point discrimination threshold at the foot sole of able-bodied subjects (12 mm [11]). We based our assumptions of perception accuracy on studies that might not apply to the residual limb inside a prosthetic socket; to get more accurate insights, studies on the force perception accuracy of amputees in their stump would be necessary. We did not analyze the swing phase; especially for prostheses that are only stable when the knee joint is fully extended the knee angle might be a useful cue.

Conclusion

We have developed a model that allows us to estimate how accurately amputees can estimate the CoP under their prosthetic foot based on observations from the literature. The resulting uncertainty in CoP perception is substantially higher (27 mm) compared to the two-point discrimination threshold of able-bodied subjects standing still (12 mm). This implies that even if they had appropriate measures of actuation, amputees would be unable to achieve similar sensing performance as able-bodied people, which could be a reason for limited balance control. Therefore, a sensory substitution system providing CoP information could indeed be useful. If this ultimately leads to improved balance in amputees will be addressed in future work.

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