Secured Microprocessor-Controlled Prosthetic Leg for Elderly Amputees: Preliminary Results

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Secure microprocessor-controlled prosthetic leg for elderly amputees: Preliminary results

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Abstract. We introduce a new prosthetic leg design, adapted to elderly trans-femoral amputees. Technical progress in prosthesis design mainly concerns active individuals. An important number of elderly amputees are not very mobile, tire easily, present reduced muscle strength, and have difficulties managing their balance. Therefore, the needs and characteristics of this specific population are very different from those of younger ones and the prosthetic solutions are not adapted. Our artificial knee has been designed to fulfill the specific requirements of this population in terms of capabilities, transfer assistance, security, intuitiveness, simplicity of use, and types of physical activity to be performed. We particularly focused our efforts on ensuring safe and secure stand-to-sit transfers. We developed an approach to control the different states of the prosthetic joint (blocked, free, resistant), associated with different physical activities. Amputee posture and motion are observed through a single multi-axis force sensor embedded in the prosthesis. The patient behaves naturally, while the controller analyses his movements in order to detect his intention to sit down. The detection algorithm is based on a reference pattern, calibrated individually, to which the sensor data are compared, and submitted to a set of tests allowing the discrimination of the intention to sit down from other activities. Preliminary validation of the system has been performed in order to verify the applicability of the prosthesis to different tasks: walking, standing, sitting down, standing up, picking up an object from a chair, slope and stair climbing.

Keywords: Prosthetic, intention detection, postural tasks, stand-to-sit

1. Introduction

Amputation of the lower limb is a growing problem in the elderly population. Indeed, the number of elderly persons is increasing, together with their susceptibility to atherosclerotic troubles. As an example, in France, which is typical of all industrial countries, 4500 patients every year are equipped, with femoral prostheses (mid-thigh). Eighty-five percentage of these patients come from the geriatric population. This number will increase because of population ageing, the increase in life expectancy, and the increase of cardiovascular pathologies such as diabetes or arthritis. A significant number of amputees over the age of 65 will be prescribed lower limb prostheses for walking. However, many amputees do not recover a high functional level after prosthetic rehabilitation [4]. The context of our article is prosthetic rehabilitation, the goal being to compensate for the loss of an amputated lower limb in order to encourage activity and achieve a level of autonomy as close as possible to that prior to amputation. The elderly, because of difficulties in walking, may restrict the use of their prosthesis, and sometimes stop using it. The loss of a limb has important repercussions on the physical, psychological and social aspects

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of the amputee's life. An efficient prosthesis should bring increasing autonomy and quality of life to the elderly by offering them the possibility of walking and standing easily and securely [3, 10]. The elderly amputee population is complex because of the normal physiological changes due to aging, such as decrease in muscle strength and bone density, associated with balance or coordination troubles. An overuse of the contralateral healthy leg is not suitable. The muscular capital is critical in actuating a prosthesis using the stump and/or crutches. Furthermore, there is a strong likelihood that this population may present other diseases (stroke, arthritis, impairment in hearing or vision, heart disease). Cognitive skills may also be decreased. The specificities of aged patients must be taken into account to evaluate the capacities available for the use of a prosthesis [11].

Today's solutions for the population considered consist mainly of mechanical prostheses equipped with a lockable knee, in combination with walking assistance devices, such as crutches or a walker [10]. During walking the knee remains locked. When wishing to sit down, the patient must manipulate a control located in his pocket or next to the prosthetic knee to release the lock. The 1M10 from Proteor is a good example of these solutions (Fig. 1, left picture).

These prostheses consist of a hinge joint for the knee, equipped with a very simple mechanical lock (a latch sliding in a slot). Positive aspects of these solutions are their low cost, lightness and robustness. But drawbacks are numerous: for example when wishing to sit down, the patient has to drop his crutches or walker to access the lock, which may affect balance. Furthermore, to unlock properly, the prosthetic leg must be partially unloaded. As a consequence, the patient has to transfer part of his weight onto his valid limb. Once unlocked, the knee is totally free to fold, and the amputee has to counterbalance his weight with his valid leg and his arms until seated.

Other prostheses exist with a friction brake, where the patient has to maintain the load on the prosthetic knee to keep it locked. The Ottobock 3R90 system belongs to this family. We consider these solutions insufficiently secure for the population we address; furthermore, users must learn specific behavior to control such devices.

A final category of knee prostheses to be distinguished is that of microprocessor-controlled prosthetic legs. These detect the intent of the patient to sit down and release the knee appropriately. The famous Ottobock C-leg belongs to this category [6, 9]. Such prostheses were initially designed for people in good physical shape, active and managing their balance with no difficulty. As a consequence, locking is not perfect, and these prostheses fold slowly, but irremediably, if load is exerted on the artificial limb. Furthermore, they are intended to flex during the swing phase of walking in order to increase gait aesthetics and efficiency, which can make them insufficiently secure for some elderly people. The C-leg Compact (Fig. 1, middle picture) is a

		
1M10 (Proteor): mechanically lockable	C-Leg Compact (Ottobock): micro processor-controlled	Our prototype: secure, micro processor-controlled, for elderly amputees

Fig. 1. Examples of prosthetic knees.

variant of the C-leg which addresses the elderly population. However, it is mechanically similar to the C-leg, and therefore not secure enough for elderly people, as the locking sequence is not perfect for walking. Furthermore, its price remains high compared to purely mechanical options. These prostheses are equipped with force (or strength) sensors and encoders, whose data are processed to estimate the patient's intent. The control algorithm is based on a state machine, where transitions between states are detected in real-time on the basis of raw sensor signals [5]. This approach is also the one selected for our application. However, control of the above micro-processor prostheses relies on the detection of numerous states and transitions. If this offers a number of possibilities for the leg, it also penalizes safety, as the intent-detection algorithm must distinguish many states one from another. Reducing the number of states and transitions, to concentrate only on the essential ones, was the way we chose to increase safety and robustness. Additionally, control of the micro-processor prostheses that fold during walking must identify the exact sequence and duration of the step in order to unlock the prosthetic knee appropriately [2]. This often takes a few steps, during which the prosthesis may react inaccurately.

Based on these considerations, we decided to design a specific prosthetic leg dedicated to the geriatric population that remains stretched when walking, thus requiring no adaptation sequence. What is proposed here is a compromise between the previously described solutions (Fig. 1, right picture): 1) a micro-processor-driven leg prosthesis that detects the patient's intent to sit down and releases the knee appropriately, 2) a knee which remains mechanically locked the rest of the time, 3) a knee which offers mechanical assistance when sitting by resisting against motion, without being totally free, for safety reasons. The novelty of our proposal is to offer a mechanical design with the intrinsic security properties associated with automatic stand-to-sit task detection. Preliminary feasibility validations have been carried out to evaluate the applicability of the prosthesis prototype.

2. Description of the system: Mechanical and functional choices

Observing the elderly trans-femoral amputee population led us to the following conclusions: some patients encounter difficulties in managing their balance and get tired quickly, so cannot stand or walk for long periods and therefore need to perform frequent sit-to-stand and stand-to-sit transfers. As a consequence, their physical activity mainly consists in making a few steps in a flat and secure environment, and achieving these transfers. Additionally, utilization of the system should be simple and intuitive as their cognitive skills are not optimal. From these observations, requirements for the prosthesis were established. Safety is one of the key factors, as, from the psychological point of view, these people must feel secure. Knee release should not require specific behavior from the patient; it should be natural and intuitive. Assistance during transfers should be provided to relieve the healthy limb. Walking aesthetics do not really matter.

We focused our attention on designing a new controllable prosthetic knee, specially adapted to non-active elderly people. This preliminary prototype offers intuitive locking and release phases, and assistance during stand-to-sit transfers. It remains locked during walking.

2.1. Mechanical design

A prosthetic leg is basically made of (i) a prosthetic foot, (ii) a prosthetic knee, and (iii) an interface with the stump of the amputated limb. We decided to select commercially available foot and interface, and to concentrate on the design of the knee only.

The knee was designed to satisfy the following functional requirements: (i) capability to flex; (ii) lockable; (iii) resisting when flexing. Design was also based on certain strategic choices such as: (iv) dissipative actuation only (knee can only resist against motion); (v) electrical control; and (vi) ability to unfold regardless of the control signal. This is to mechanically bypass the electronics and allow the user to rise even in the case of control algorithm malfunction.

These aspects will now be discussed in detail:

2.1.1. Knee flexion

In existing prostheses, two types of joint giving the knee the capability to flex can be distinguished: four-axis knee joints that mimic human knee kinematics (such as on the 1M05 from Proteor), and simple hinge-joint knees (such as on the 1M10 from Proteor). The first category is more secure than the second, as the instantaneous centre of rotation is located behind the leg when stretched, and the weight of the patient tends to push the knee against its mechanical stop, thus maintaining it stretched. For reasons of simplicity, and because the control is supposed to maintain the knee locked when walking, we decided to use a simple hinge joint for the knee.

2.1.2. Total locking of the knee when the leg is stretched

Giving the prosthesis the ability to remain perfectly locked during knee extension was an important safety concern.. However, it was observed that for some robotized knee prostheses (such as the C-Leg Compact from Ottobock) only partial locking was ensured. A strategy to perfectly lock the knee was preferred, considering the population targeted.

2.1.3. Resistance during flexion

One specificity of our knee is to assist the patient during stand-to-sit transfers, while allowing him to share his weight between valid and prosthetic legs. However, classical prostheses for elderly people, made with a mechanically lockable hinge joint, are totally free to fold once unlocked. This obliges the patient to sustain all his weight with his valid limb.

2.1.4. Dissipative actuation

Dissipative actuation was chosen to lower energy consumption, and to offer the device more autonomy. Dissipative actuation means that the knee can only resist against motion. Hence, it can efficiently assist the patient during the stand-to-sit transfer, but is unable to provide torque when the patient is rising. However, although passive, the knee prosthesis we propose is designed to assist the patient while rising, by showing no resistance when unfolding, and being able to relock automatically if the patient stops rising. Again, this allows the patient to take his weight on both legs, and promotes a feeling of safety.

2.1.5. Electrical control

We opted for electrical control of the knee, using a microprocessor and sensors. This choice was made because sensor-based control is more efficient (more secure and more easily tuneable) than control via fully mechanical systems (such as the mechanical knees that lock and release automatically depending on the load exerted on the prosthesis; e.g. 3R90 from Ottobock).

2.1.6. Ability to extend regardless the control signal

This choice was made to simplify the intent-detection algorithm and to concentrate on the stand-to-sit transfer only. Hence, when wishing to stand up, the patient can achieve knee extension freely. This choice allows the intent-detection algorithm to be very simple (and robust), as only the intent to sit down need be recognized.

A first prototype was designed to fulfil the needs listed above (see Fig. 2). It consists of an articulated frame with a hydraulic cylinder and an electro-valve (Fig. 3).

A micro-electro-valve, switched on and off by the microcontroller, allows oil to go from one side of the cylinder to the other, or not. It is mono-stable, normally locked, thus requiring no energy to remain locked. Additionally, it has no leakage when closed, which ensures total locking of the knee. The cylinder is connected to the knee joint with classical pivot joints. This system, taken from classical cylinder prostheses, delivers non-constant resisting torque as the lever arm of the cylinder moves while the prostheses folds: it means that, to resist against a given knee torque, the cylinder force must increase while the piston is pushed in. In order to provide a relatively constant resistive torque, the cylinder was equipped with multiple orifices, that gradually increase the resistive force by resisting against the fluid flow more and more. A flow-adjustable needle was also added to

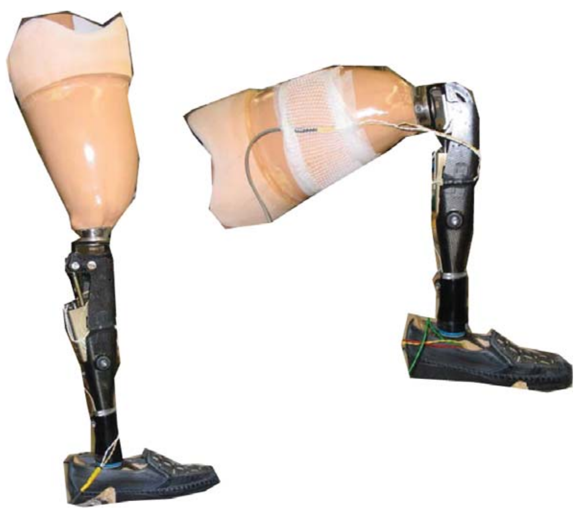

Fig. 2. The constructed prototype of the knee prosthesis.

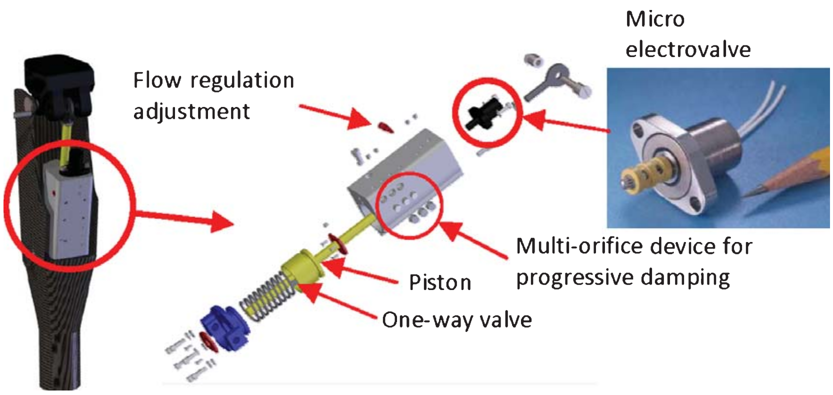

Fig. 3. Exploded view of the prosthetic leg.

allow resistance adjustment. The piston was equipped with a one-way valve, to allow the knee joint to unfold regardless of control (even with the electro-valve locked). Hence, when the knee unfolds, the resistive orifices are short circuited, and the cylinder offers very low resistance. The one-way valve was chosen to have almost no leakage when blocked, to show very low resistance when open, and to fit the piston.

The main characteristics of the prototype are: (i) total weight (with all except the prosthetics foot and the interface connecting the patient stump) is 1.380 Kg, the frame being made of carbon fibre and the cylinder of aluminium, (ii) Flexion torque before starting to leak for the cylinder is about 80 N.m, (iii) size can be adjusted to fit a standard human adult.

The hydraulic cylinder appeared to be the best option at this time, as we found a commercially-available electro-valve which was compact, light, reliable, able to totally block the fluid flow, and controlled by a very low-energy electrical signal (from the Lee company). This allowed us to shorten the design stage and the tuning phase. It is to be noted that other technologies, such as mechanical brakes or magneto-rheological clutches, were also considered. Using magneto-rheological clutches was a solution we rejected, as total locking cannot be achieved easily. Using a mechanical brake might have been a good solution, being cheaper than the hydraulic option. However, as no commercially-available mechanical brakes of that range of torque, size and weight exist, we would have had to develop a new mechanical brake starting from scratch, which would have demanded a lot of time for design and tuning.

2.2. *Instrumentation*

To make the first tests we selected a quite exhaustive set of sensors such as (i) foot-pressure sensors, (ii) accelerometers and gyrometers, (iii) a knee-angle encoder. This equipment helped us to gain knowledge of the behaviour of the elderly, especially in the stand-to-sit transfer. After carrying out these initial experiments with patients, we concluded that a single force/torque sensor would be sufficient for intent-detection (see next section). This solution allows the embedding of all instrumentation in the prosthesis, making the system much more acceptable. A very small force/torque sensor from ATI was selected. We showed experimentally that the three components of the saggital plane of the measured wrench were sufficient to detect stand-to-sit intention. That is why we are now considering embedding three mono-axial force sensors that would provide the required saggital components of the wrench, rather than this 6-axis sensor.

2.3. *Control hardware*

Even if the final goal for the prosthesis is to be totally autonomous, with embedded micro-controller and batteries, an external PC was preferred to simplify development at this stage of validation. It was equipped with an I/O board connected by cable to the sensors and the electro-valve. Embedding all the electronics will present no difficulty. We would like to insist here on the objective of the prototype developed: the aim was to validate our functional choices for the artificial knee.

The final technological design will only be developed once this preliminary phase is completed.

The leg was also equipped with a vibrating system, similar to the ones embedded in mobile phones. This warns the patient discreetly that his prosthetic knee is unlocked.

3. Exhibition and detection of movement signatures

Human movements are highly repetitive. One hypothesis to explain this is the existence of a reduced number of variables which are controlled continuously by the Central Nervous System (CNS). This simplification of the problem to be solved is expressed through close relationships between the movements involved in any given task, which allows the number of degrees of freedom simultaneously controlled to be reduced. The variables controlled are associated with low variability [7, 8]. On this basis, we formulated the assumption that the observation of these controlled degrees of freedom, due to their theoretically low variability, should allow us to define reliable signatures of postural activities. We have validated this approach for sit-to-stand tasks [1] and extended this to stand-to-sit in elderly trans-femoral amputees. The principle of the intent-detection approach is: 1) off-line, to identify a reference pattern which is relevant, and characteristic of the postural task considered, 2) on-line, to observe sensor data and recognize the reference pattern. The difficulty is to elicit a reference pattern which corresponds to the signature of the preparation of the movement to be detected. This pattern will be associated with the intent of the patient to achieve the postural task in question. The reference pattern constitutes an indirect measurement of the observed behavior.

3.1. Existence of invariant patterns in amputees' stand-to-sit transfers

We performed preliminary experiments with two elderly amputees in order to analyze the stand-to-sit transfer: we recorded the dynamic efforts applied to the ground via instrumented insoles (3 contact points recording normal forces). 3D acceleration and orientation of the trunk were also measured using a sensor placed at the C7 cervical level. The same type of sensor was placed on the valid leg (Fig. 4). The objective was to validate that patients have repetitive behavior when performing the stand-to-sit. And indeed, this first experimental campaign allowed us to observe low variability in the data measured from one trial to another for each individual (Fig. 5). This confirmed the existence of individual repetitive patterns associated with the transfers. The next step was then to define variables which we could rely on to represent the stand-to-sit transfer for different patients.

3.2. Reference pattern definition

From the previous experimental results, for practical reasons and to be acceptable in a rehabilitation context, we limited ourselves to observing these movement invariants indirectly, via dedicated sensors embedded in the prosthetic limb only. The prosthesis prototype was therefore equipped with a force/torque sensor to measure its interaction with the outside world. A

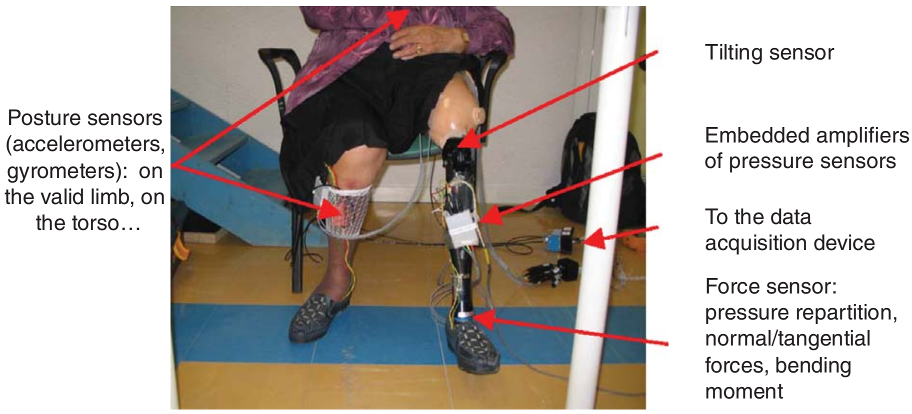

Fig. 4. One patient equipped with the prosthetic knee and motion sensors.

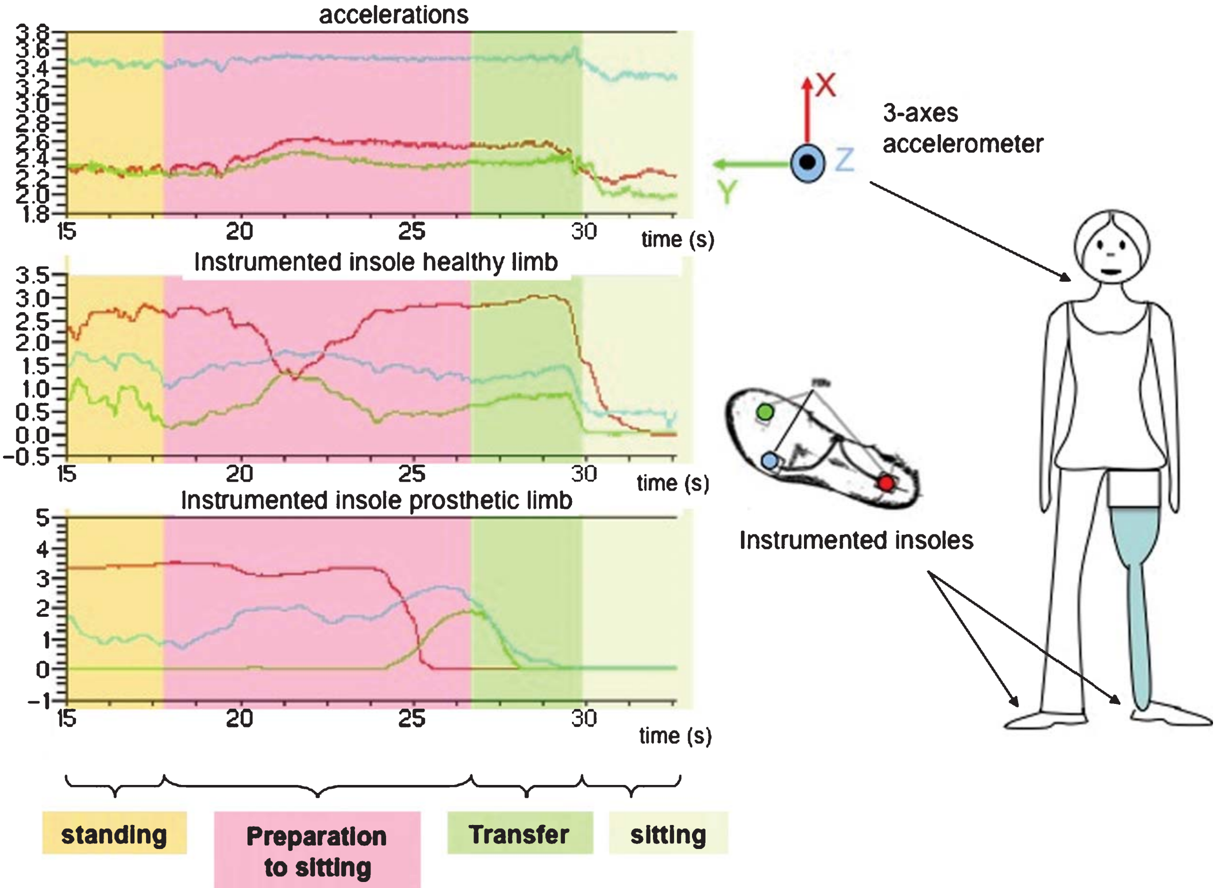

Fig. 5. Analysis of one amputee stand-to-sit transfer (1 trial).

very compact 6-axis force/torque sensor from ATI was selected and placed next to the ankle of the artificial limb.

To define a reference pattern, work was performed with able-bodied subjects to simplify the experimental protocol. We designed an adaptor to fit able-bodied subjects with the prosthesis (Fig. 6). The protocol consisted in: from a standing posture, straightforward walking over 5 meters, and sitting on a chair (the able-bodied subject was asked to sit on the side of the chair, to avoid the adaptor colliding with the chair). An experimenter using a manual switch connected to the prosthesis via a cable controlled the unlocking of the knee. Each individual carried out 5 trials. We equipped 5 subjects and recorded and analyzed data from the sensor.

Recordings clearly indicated that the main data to be considered were the sagittal plane components $(F_x, F_z, M_y^{\mathrm{O}})$ of the wrench $(\boldsymbol{R}, \boldsymbol{M}^{\mathrm{O}})$, as the other values remained very low during the stand-to-sit transfer (see Figs 7–9, F_y, M_x^{O} and M_z^{O} values):

$$\boldsymbol{R} \simeq \begin{bmatrix} F_x & 0 & F_z \end{bmatrix}^{\mathrm{T}}, \quad \boldsymbol{M}^{\mathrm{O}} \simeq \begin{bmatrix} 0 & M_y^{\mathrm{O}} & 0 \end{bmatrix}^{\mathrm{T}} \qquad (1)$$

Processing the raw signals led us to pertinent signals regarding the postural task dynamics. Intuitively, during the stand-to-sit transfer, the patient needs to apply sufficient force to his stump for the prosthetic knee to flex. Therefore, pertinent data for establishing a reference pattern might be the torque at knee level, the position of the centre of pressure, the angle of the ground reaction force and its intensity.

The knee torque M_y^{K} is computed straightforwardly using the rigid body transformation:

$$M_y^{\mathrm{K}} = M_y^{\mathrm{O}} + (\mathbf{KO} \wedge \boldsymbol{R}) \cdot \boldsymbol{e}_y = M_y^{\mathrm{O}} + \mathrm{KO}_z \; F_x \qquad (2)$$

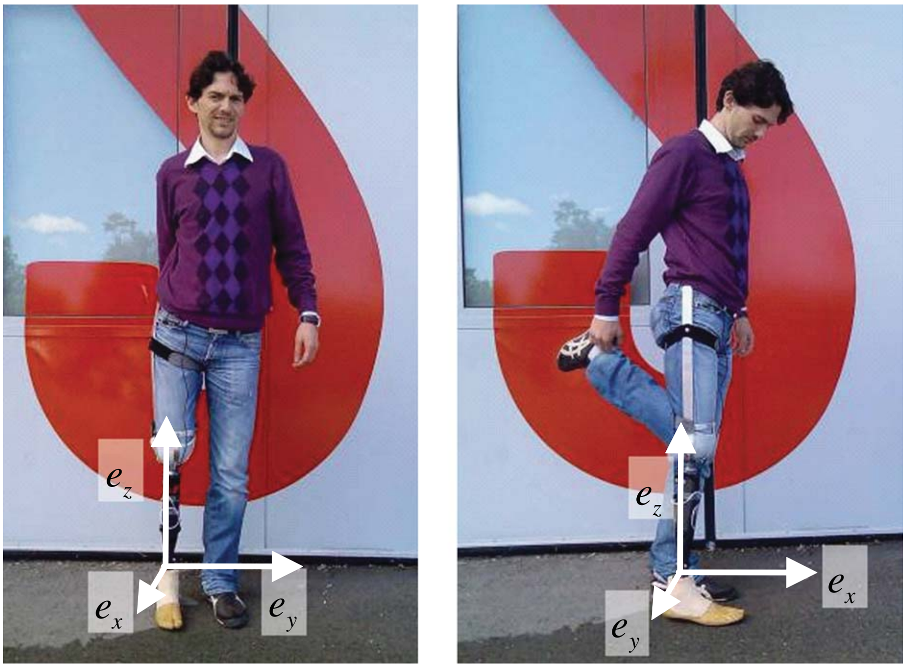

Fig. 6. Able-bodied subject wearing the prosthesis.

where $KO = [0 \quad 0 \quad KO_z]^T$ is the vector between points K and O (K representing the knee centre position and O the sensor centre position). $\wedge$ represents the cross product and $\cdot$ the scalar product.

As the problem is planar, the position of the Centre of Pressure C is computed using the values corresponding to the position on the sole of the foot where the ground reaction wrench momentum vanishes:

$$\begin{aligned} M_y^C &= M_y^O + (\mathbf{CO} \wedge \boldsymbol{R}) \cdot \boldsymbol{e}_y \\ &= M_y^O + x \ F_z - z \ F_x = 0 \end{aligned} \tag{3}$$

with $\mathbf{CO} = \begin{bmatrix} x & 0 & z \end{bmatrix}^T$ being the vector from C to O. x, the required unknown, represents the position of the center of pressure along $\boldsymbol{e}_x$. z is the distance from the sensor to the floor level measured along $\boldsymbol{e}_z$.

This leads to:

$$x = (F_x z - M_y^O)/F_z \tag{4}$$

Note that when $F_z = 0$, x cannot be established. This corresponds to the fact that, when the foot is in the air, no vertical force acting on the foot is recorded, and the centre of pressure is undefined.

The angle α of the ground reaction force is given by:

$$\alpha = \tan^{-1}(F_x/F_z) \tag{5}$$

and the norm R of the ground reaction force by:

$$R = \sqrt{{F_x}^2 + {F_z}^2} \tag{6}$$

Analyzing these plots (see below figures) showed that knee torque M_y^K appeared to be an interesting reference variable. The correlation of this variable with knee unlocking is clear. Its low intra-variability in terms of absolute values is interesting for calibration procedure, while the repetition of the pattern shape between individuals indicates that a generic algorithm may be used to detect the pattern.

3.3. Detection algorithm

From the exhibited reference pattern M_y^K we designed a stand-to-sit detection algorithm, based on several tests, to be validated for knee unlocking and to guarantee the security of the transfer.

The algorithm is based on a succession of tests in order to determine from the patient's position whether the patient is about to sit down, and whether the con-

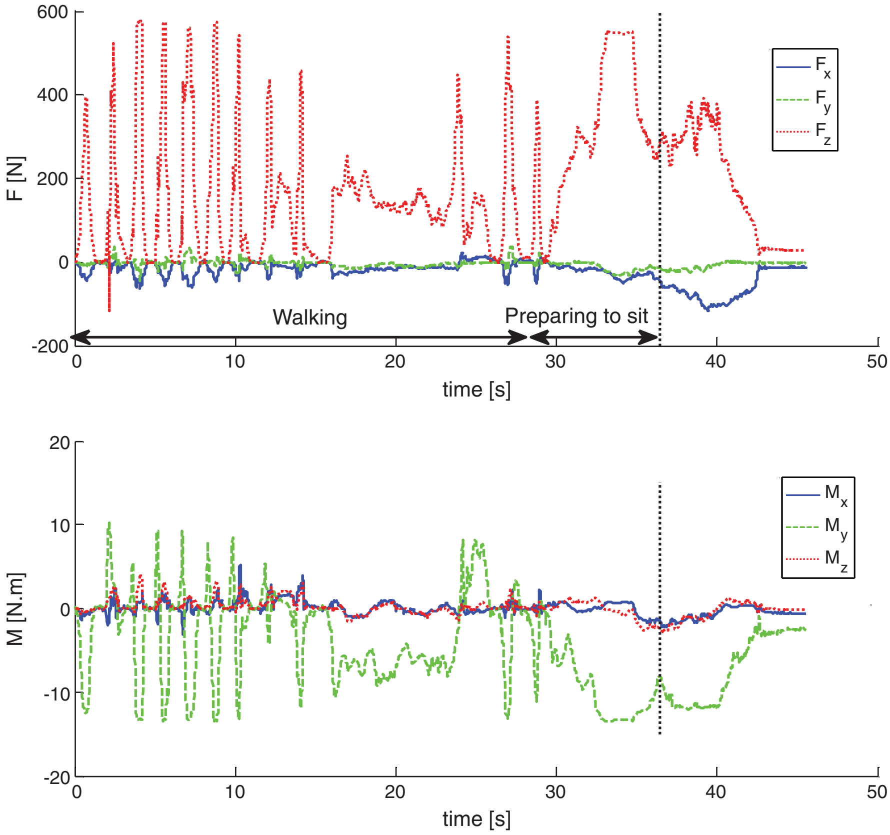

Fig. 7. Ankle force/torque sensor raw data (vertical lines indicate the instant of knee unlocking).

ditions are sufficiently secure to allow the knee to unlock.

1) The patient applies effort on his legs in order to bend them. In other words, M_y^{K} increases significantly and is greater than a threshold value which needs to be calibrated. (See Table 1 for numerical values.) This test can be mathematically expressed as follows:

$$\text{TEST 1}: \quad M_y^{\mathrm{K}} > M_y^{\mathrm{K}} \tag{7}$$

2) Before sitting, the patient must be standing quietly. In other terms, no impacts must have been detected over a given time horizon (to be calibrated). This verification is performed via analysis of the resultant force.

Let mean(R, n) represent the mean value of R over n samples:

$$\mathrm{mean}(R, n) = \sum_{i=0}^{n-1} R[k-i]/n \tag{8}$$

where k is the current sample corresponding to the current time t $(t = k/f_{\mathrm{acqui}})$ with f_{acqui} being the acquisition frequency (during our experiments: $f_{\mathrm{acqui}} = 100\,\mathrm{Hz}$).

This test is expressed by:

$$\text{TEST 2}: \quad |\mathrm{mean}(R, n) - R[k]| < \Delta R_{\mathrm{threshold}} \tag{9}$$

with $\Delta R_{\mathrm{threshold}}$ the threshold for the force resultant variation and n corresponding to τ $(n = \tau \times f_{\mathrm{acqui}})$ the time during which the patient must be standing quietly.

3) For safety reasons and quality of use, the patient's weight must be distributed between the two feet. This distribution is adjusted and should be modified once the patient gets used to his prosthesis (sufficient training may permit equal distribution).

$$\text{TEST 3}: \quad F_{z\ \mathrm{threshold\_min}} < F_z < F_{z\ \mathrm{threshold\_max}} \tag{10}$$

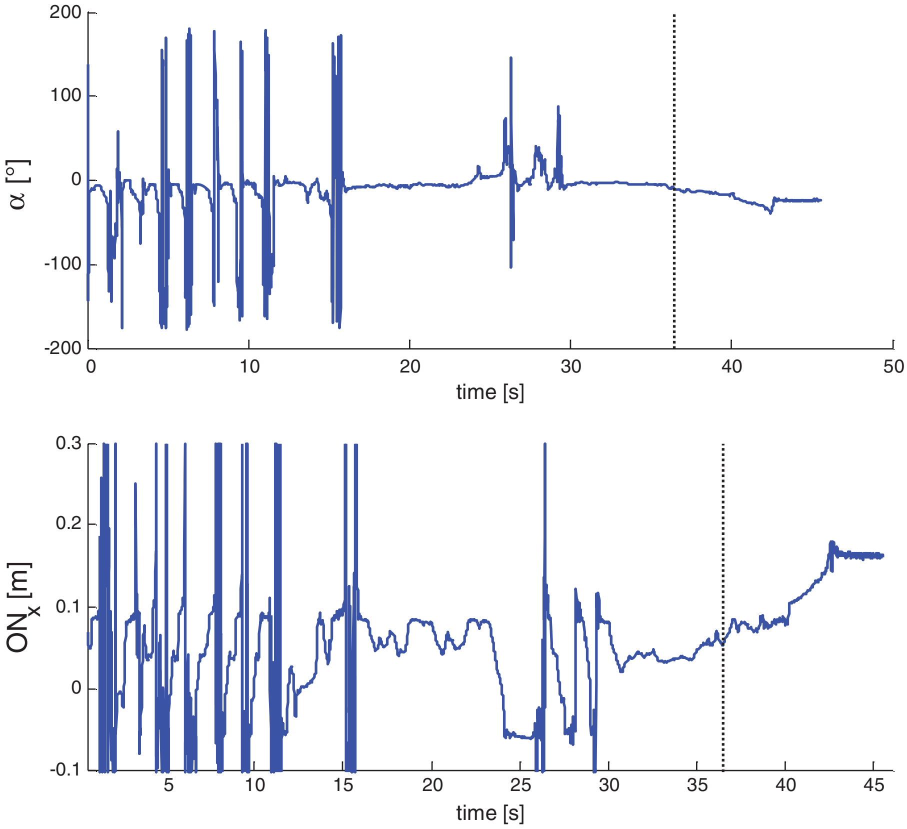

Fig. 8. Ground/foot force angle and centre of pressure position (vertical lines indicate the instant of knee unlocking).

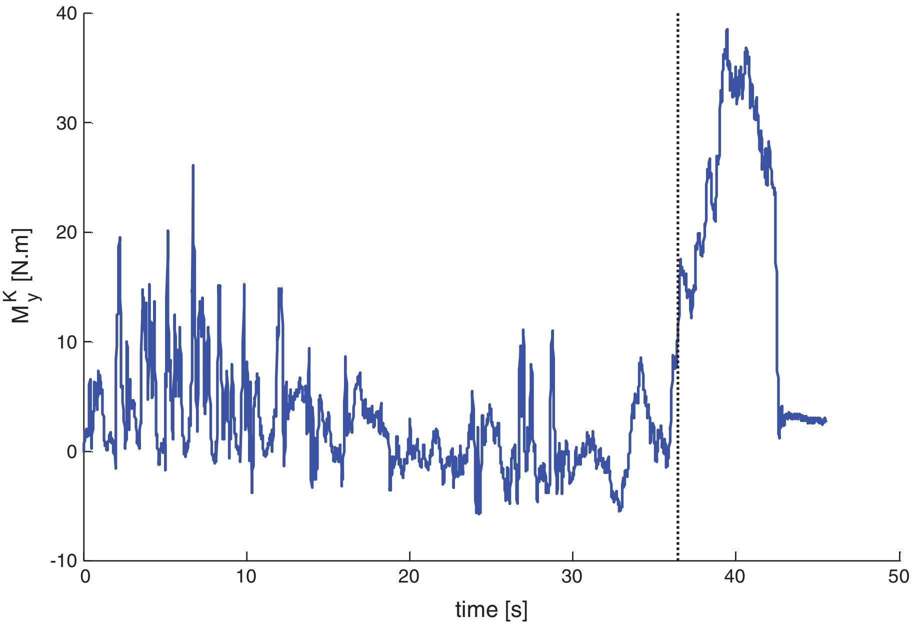

Fig. 9. Knee moment (vertical lines indicate the instant of knee unlocking).

4) This final test was added later. It consists in checking that the patient is in a safe position. The trunk should be slightly bent forward so that the patient does not fall backwards when the knee unlocks. This should allow the intent-detection algorithm to differentiate a stand-to-sit intent from descending a slope (see below). We expressed this via the position of the center of pressure in relation to a threshold to be calibrated

Table 1
Test threshold values used for patient

Knee moment threshold	M_{yK}	20 N.m
Time window duration	τ	3 s
Resultant variation threshold	$\Delta R_{\text{threshold}}$	100 N
Vertical force min threshold	$F_{z\ \text{threshold\_min}}$	250 N
Vertical force max threshold	$F_{z\ \text{threshold\_max}}$	400 N
Center of pressure threshold	$x_{\text{threshold}}$	0.08 m

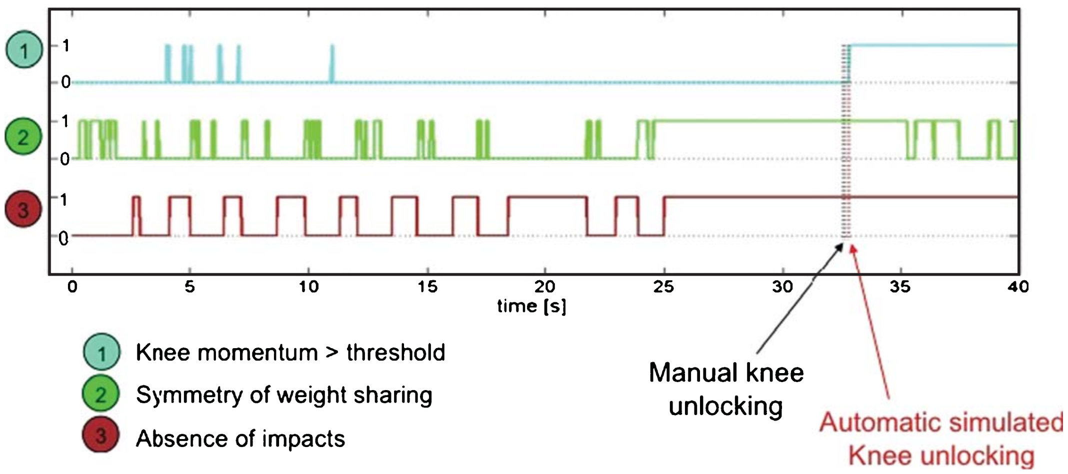

Fig. 10. Performance of the detection algorithm tested on one trial from one able-bodied subject (walking → quiet standing → sitting).

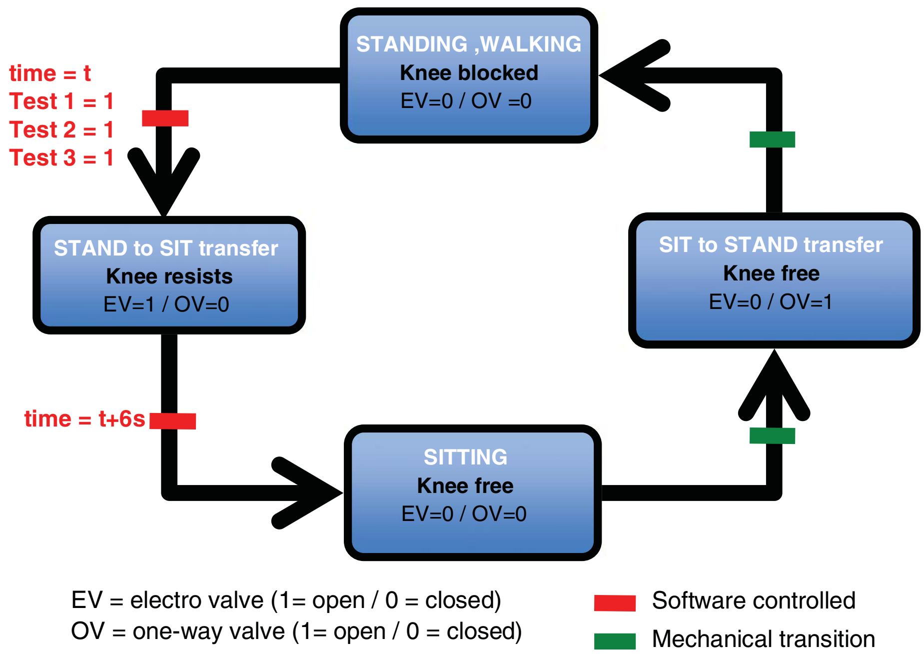

Fig. 11. States and transitions scheme of the intent detection algorithm.

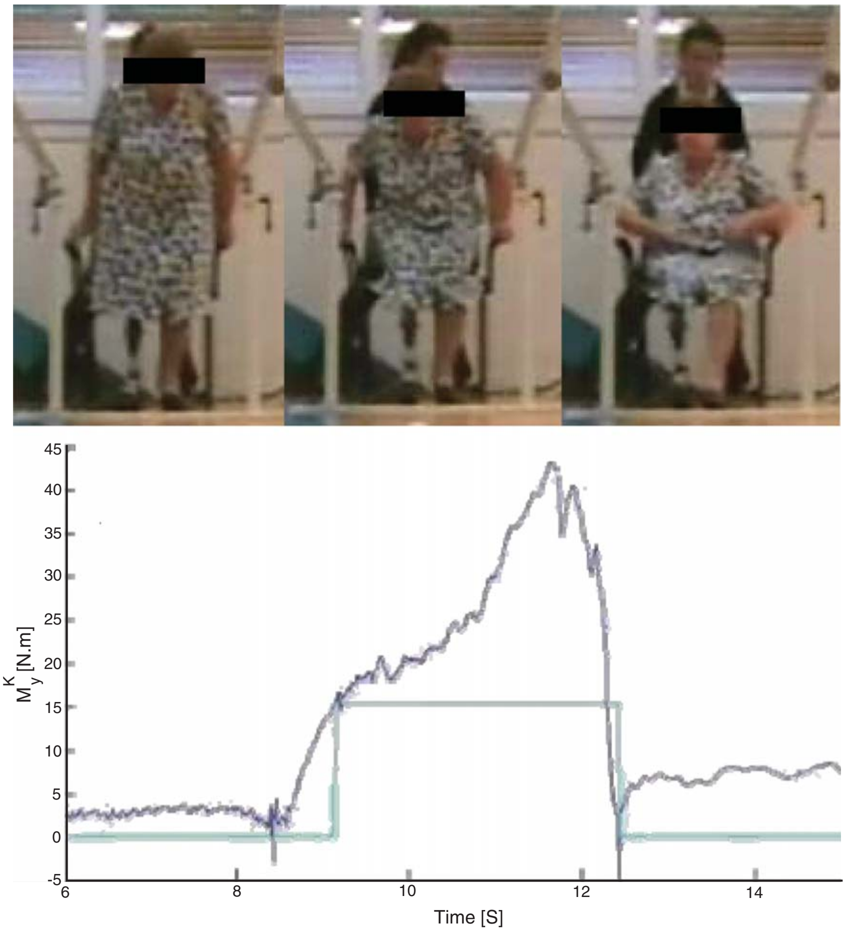

Fig. 12. Stand-to-sit task performed by an amputee patient wearing our prosthesis. Top: Pictures from video recording – Bottom: Knee moment evolution in dark blue.

individually, depending on the patient's size and weight.

$$\text{TEST 4}: \quad x > x_{\text{threshold}} \tag{11}$$

We illustrated this test algorithm (T1 to T3) on data recorded from one able-bodied subject in Fig. 10.

Data recorded during one trial were used to simulate knee-unlocking control via the detection algorithm. The instant of unlocking can be compared to the actual unlock controlled manually by the experimenter.

An efficient calibration of the thresholds allows the generation of a simulated unlocking command very close to the real one.

4. Experimental validation

One amputee patient (female, 65 years old, 1.60m, 80 kg) was equipped with our prosthesis. The knee was connected to a PC.

The protocol consisted in a first series of stand-to-sit trials in order for the patient to get used to the prosthesis. The experimenter unlocked the knee manually when the posture of the patient appeared correct for knee flexion.

The final trials were used to calibrate the detection algorithm in terms of threshold values, according to the procedure presented in the previous section.

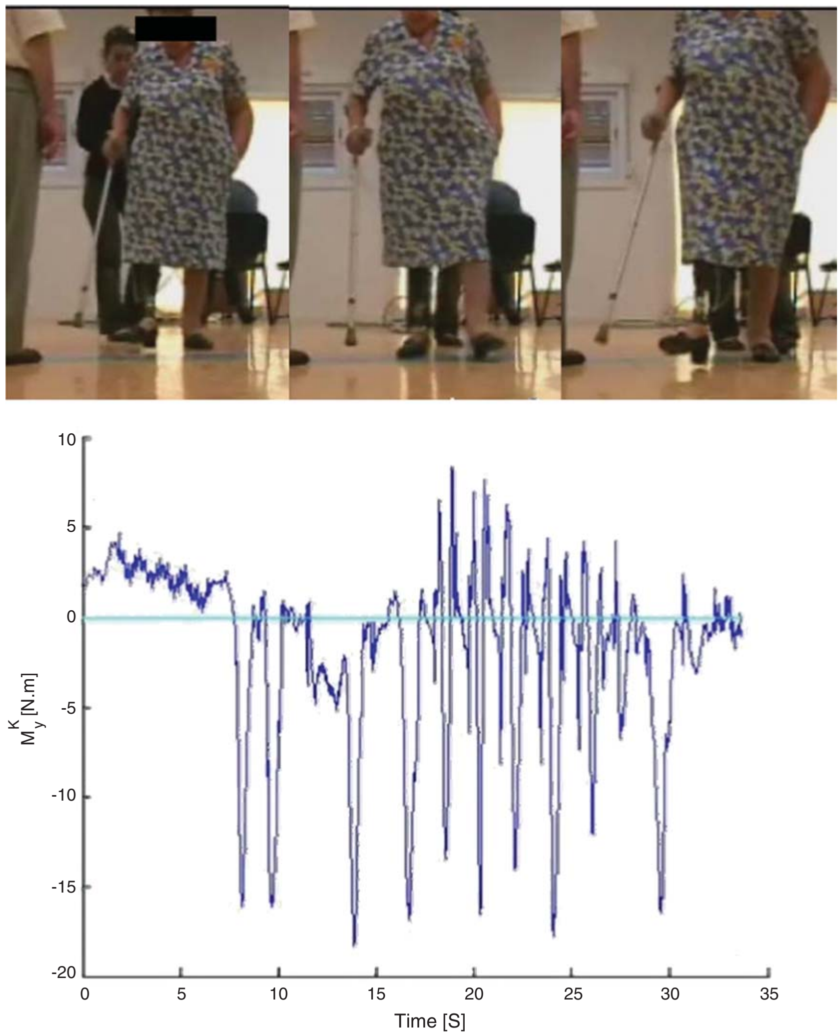

Fig. 13. Walking task performed by an amputee patient wearing our prosthesis. Top: Pictures from video recording, Bottom: Knee moment evolution.

From this point, a PC controlled the knee joint using the detection algorithm.

The patient was asked to stand-to-sit (Fig. 12), sit-to-stand, walk on uneven ground (Fig. 13), climb stairs, walk up and down a slope, and take an object placed on a chair.

In all the trials, the stand-to-sit, sit-to-stand and walking were perfectly discriminated and the joint unlocked adequately. No problem was found concerning walking up a slope, object grasping or stair climbing.

The patient walked down the slope correctly, but the data analysis showed errors in the detection algorithm, which detected "sitting intention". This is mainly due to the fact that the displacement is very slow and high torques are applied to the knee during this type of task. Test 4 proposed in our detection algorithm should allow correct discrimination, but this has not yet been experimented.

The patient was very enthusiastic and reported that the system was comfortable, intuitive, and easy to use.

5. Conclusion and perspectives

This paper presents an original, secure, robotized prosthetic knee dedicated to elderly femoral amputees. We focused on assisting stand-to-sit transfers in a safe and secure manner. The artificial limb controller is able to detect the patient's intention to sit in order to unlock the joint. It is also perfectly secure during walking. The principle of the detection algorithm is based on the observation of patient movements and posture, and on an individually-calibrated reference pattern. Intra-variability in postural movements is very low, which allows high predictability and robustness of motion signatures. Independently of the sensor choice, the principle of the algorithm is to identify, in the measured signals, the reference pattern preceding the transfer from standing to sitting. This detection allows knee unlocking to be initiated. Detection must be integrated as soon as the task is initiated in order for the patient to sit as naturally as possible and to use his leg instinctively.

Discrimination from other postural tasks was validated, except for walking down a slope, which will need additional analysis and should lead to new tests on our algorithm.

We have chosen to use a detection based on tests, but other approaches may be envisioned: correlation, abrupt change detection, event classifiers, discrete step machines or neural networks. If additional states of the prosthesis are proposed in a later generation, the complexity of the problem may lead us to choose other types of technique to minimize recognition errors.

Our robotized prosthetic leg is dedicated to a non-active elderly population of femoral amputees; its main advantages are: secure transfers, integrated sensors, simplicity and intuitiveness of use. The sensors now need to be changed, in order to reduce the price of the system.

Additional clinical experiments need to be carried out, to compare over a longer period the occurrence of fatigue when using our prosthetics or classical ones.

In the future the system will also be improved in order to address other postural tasks.

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