Chapter 4
Development of a Bench for Testing Leg Prosthetics

H. Giberti, F. Resta, E. Sabbioni, L. Vergani, C. Colombo, G. Verni, and E. Boccafogli

Abstract The present paper deals with the development of a bench for testing prostheses for legs. The bench includes a fake leg (from femur to foot) and it allows rotations of femur (up to 40°) and of tibia with respect to femur (up to 40°). Lockable gas springs allow stopping tibia and femur rotations. An hydraulic actuator applies a load to the hip, while a sled driven by an electric motor moves the foot simulating walking. Attachment between foot and sled has been designed to allow load transfer from heel to forefoot. Once maximum rotations for tibia and femur are reached, the leg is lifted and swings. As a first application, the test bench will be used to test transtibial prosthetics, including the knee socket, but it has been designed so that it could allow testing, with a limited number of adjustments, also transfemoral, foot and ankle prosthetics.

Design of test bench components and definition of control strategy for moving and synchronizing actuators, has been achieved through co-simulation of a Multi-Body model of the bench developed using ADAMS/View and a model of the control system and actuation devices developed using MatLab/Simulink.

Results of preliminary experimental tests on a transtibial prosthesis are shown.

Keywords Transtibial prosthetics • Lower-limb amputee • Test bench design • Multi-body model • Co-simulation

4.1 Introduction

The present paper deals with the development of a bench for testing prostheses for, which is carried out within a research project supported by the centre for testing and application of prosthetics and orthopaedics of INAIL (Istututo Nazionale per l'Assicurazione contro gli Infortuni sul Lavoro-Italian National institute for industrial insurance). The test bench includes a fake leg constituted by elements reproducing femur, tibia, ankle, foot, knee and knee socket (Fig. 4.1). Aim of the bench is to test prostheses considering working conditions more similar to real ones with respect to methodologies provided by standards [1, 2], although, as it will be explained later on, it doesn't completely reproduce walking. Moreover, while standards provide methodologies for testing a single prosthesis type (foot, ankle, etc.), the proposed bench would allow to contemporarily test prostheses for different parts of the leg with a few adjustments.

In this paper the design of the test bench is discussed, considering its application to transtibial prostheses used by lower-limb amputees (Fig. 4.2). The bench has therefore to guarantee the sequential replication of loads on the prosthesis and it has to reproduce relative rotations between femur and tibia during walking, also allowing the adaptability to different patients.

A series of assumptions have been made in the test bench design:

- Although the movement of the leg is a 3D motion, only the plane of walking is taken into account (sagittal plane). Loads and stresses in the other two planes (frontal and transversal) are indeed neglected, since reasonably lower. Under this assumption, the structure undergoes a bi-dimensional loading system;

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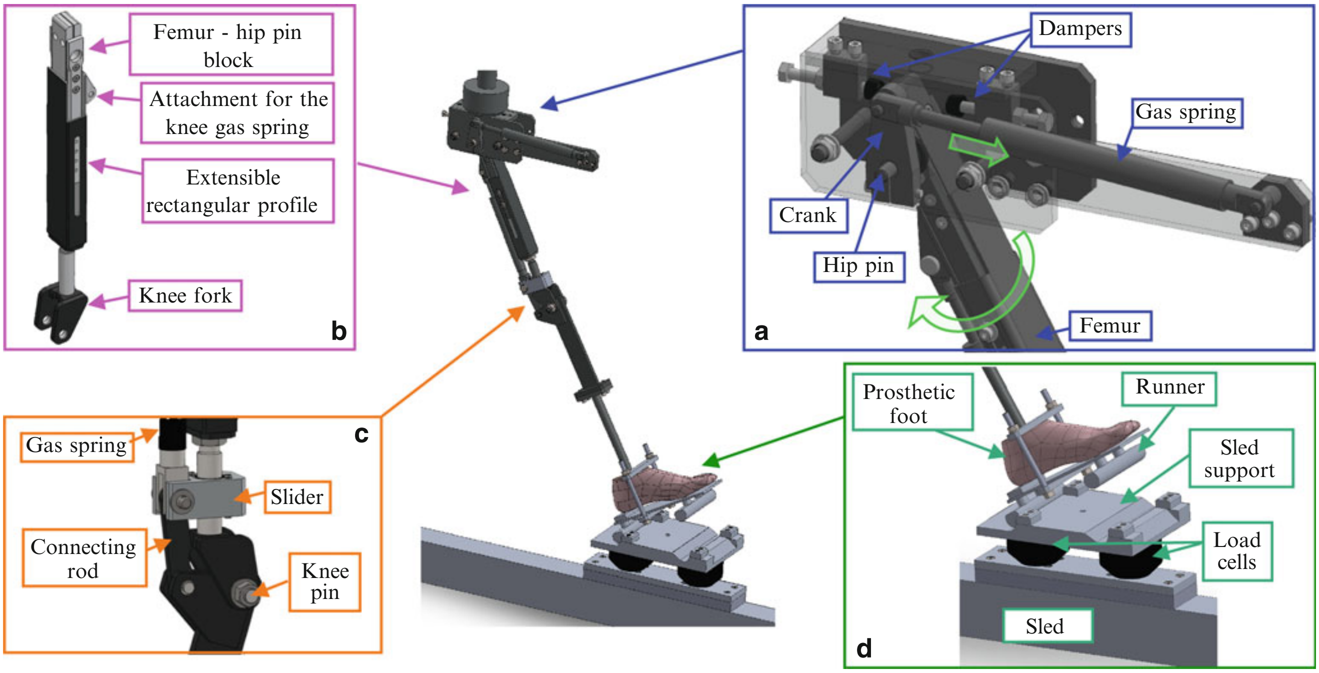
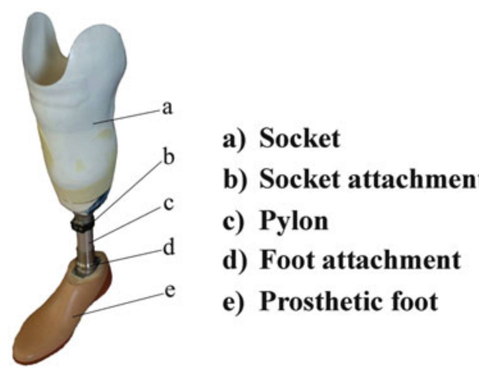


Fig. 4.1 Assembly of the test bench and some of its details: (**a**) hip assembly; (**b**) femur; (**c**) knee system; (**d**) foot and sledge joint

Fig. 4.2 Components of a transtibial prosthesis



- Despite hip and knee can be regarded as roto-transltional joints, they are introduced in the test bench as revolute joints (i.e. hinges);
- No relative rotations between tibia and foot are allowed since, generally speaking, in transtibial prostheses used by lower limb amputees, the prosthetic foot is clamped to the pylon reproducing tibia and fibula bones;
- Hip can only move along vertical direction. This assumption implies that motion of leg parts can only be achieved by imposing a motion to the foot. As it will be explained later on, motion of the hip and the load acting on it (reproducing the patient weight) are applied by an hydraulic actuator, while the foot is moved by a sled driven by an electric motor. This assumption represents a significant difference with resect real walking, even if reproduction of relative rotations between leg components is allowed;
- Rotations of femur (angle α) and of tibia respect to femur (angle γ, see Fig. 4.3) are applied one after the other, while during real walking, they take place simultaneously;
- Femur rotations (angle α, see Fig. 4.3) in the range [$-10°$; $30°$] and rotations of tibia respect to femur (angle γ, see Fig. 4.3) in the range [$0°$; $30°$] are considered (in amplitude). Although literature gait analysis [3] and experimental tests performed at INAIL labs have shown that femur rotation varies from $30°$ at the beginning of the stance phase to $-10°$ before the beginning of the swing phase, and that tibia-femur relative rotation (angle γ, see Fig. 4.3) is almost zero during the stance phase, but reaches values of approximately $60°$ in amplitude before the swing phase, largest stresses in the prosthetic components are reached during the stance phase, this justifying the selected range for angles α and γ.

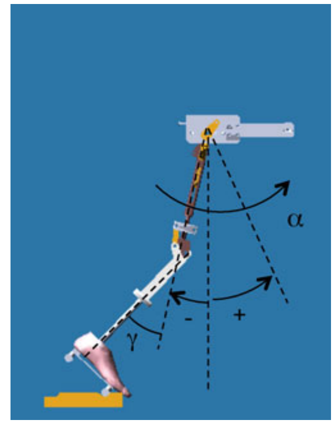


Fig. 4.3 Angles and conventions

4.2 Design of the Test Bench

Figure 4.1 shows the test bench designed under the assumptions reported in the previous section. Motion and load conditions to the fake leg are provided by an hydraulic actuator and a sled driven by an electric motor. The hydraulic actuator drives the vertical motion and the vertical load applied to the hip. The sled instead provides the foot motion.

Components of the test bench can be divided between passive and active.

Passive components are:

- A rectangular profile reproducing the femur (Fig. 4.1b). It has a variable length between 350 and 450 mm, corresponding to lengths ranging between the 50% percentile of women and 95% percentile of men;
- The knee articulation mechanism connecting femur and tibia. This mechanism must be included inside the reproduction of an amputee stump fitting the knee socket, to allow its testing. Dimensions of the mechanism must thus be limited since, in the knee region the geometry of the amputee stump decreases and the maximum allowed section is $80 \times 80\ \text{mm}^2$;
- A continuous pylon representing the tibia, bolted at the desired length;
- The prosthetic foot, bolted to the tibia;
- The attachment of the prosthetic foot to the linear sled (runner, Fig. 4.1d). The foot is clamped in a structure including a platform connected with two semi-cylinders. Contact occurs between the semi-cylinders and the sled support. This attachment is needed to allow the sled to hook and drag the foot. Geometry of attachment has partially been inspired by ISO-10328 [1]. At the beginning of gait, contact occurs only on the rear cylinder (i.e. the heel), then it moves to the front one (i.e. the forefoot). Distance between the two regions has been defined according to ISO-10328.

All the above components can be replaced with the prostheses to be tested. In particular, in the application considered in this paper, tibia is replaced by a transtibial prosthesis, including the knee socket.

Active elements of the bench needed to provide the desired motion and load conditions to the prostheses to be tested are:

- An hydraulic actuator is attached one side to the hip and the other to a fixed frame. As already mentioned, it drives the vertical motion of the hip and the load applied on it. Load is provided in agreement with ISO-22675 [2] and it depends on the leg positions during walking. A load cell measures the force provided by the hydraulic actuator, while a LVDT measures its displacement;
- A lockable gas spring between hip and femur. The air spring can be locked pressing an on-off button. An automatic system constituted by a small hydraulic piston has been designed and interfaced with the control logic to commute the air spring condition from locked to unlocked and viceversa during the work cycle. When the air spring is locked, femur rotations are prevented, while when it is unlocked femur rotations are allowed (Fig. 4.1a);
- A lockable gas spring between femur and tibia. The air spring acts on a kinematic mechanism consisting of a slider, with an axial bearing, and a connecting rod, which allows relative rotations between tibia and femur (Fig. 4.1c). This mechanism has been design to fit the morphology of the socket. In the knee region, in fact, the geometry of the amputee stump decreases and it allows only limited dimensions for the components to be used (maximum section $80 \times 80\ \text{mm}^2$). When the air spring is locked, relative rotations between knee and femur are blocked.
- A sled moving on a linear guide driven by an electric motor. Length of the linear guide is 1.5 m to reach the desired rotations of the leg parts. The linear guide has been selected on the basis of the maximum vertical load that must be applied to the leg (2,600 N during static tests). On the sled are placed two load cells placed in correspondence of the semi-cylinders of the foot platform.

All rotating elements, such as hip and knee pins, are provided with bushings.

Dimensions of mechanical components are obtained considering structural integrity both for static loads and infinite fatigue life.

To select the electric motor driving the sled, the hydraulic actuator and the gas springs and to evaluate the loads of the elements constituting the test bench for their dimensioning, a Multi-Body model of the test bench has been developed.

4.3 Model of the Tests Bench

Selection of the actuators and assessment of the capability of the designed test rig of simulating the work cycle for fatigue tests (i.e. a step) has been achieved through co-simulation of a Multi-Body (MB) model of the test bench, implemented in ADAMS/View environment, and a model of the control strategy including actuating devices (hydraulic actuator, electric motor and gas springs), implemented in MatLab/Simulink environment. The control strategy has been developed in MatLab/Simulink environment to be directly transferred on the Real-Time (RT) board driving the actuators of the test bench.

Figure 4.4 shows the mechatronic MatLab/Simulink model; the ADAMS_sys block represents the MB model of the test bench developed using ADAMS/View: inputs of the MB model are the torque provided by the electric motor driving the sled, the force provided by the hydraulic actuator moving the hip and the forces provided by the gas springs. Outputs of the MB model are the electric motor angular speed (from which the position of the sled is derived) and force and displacement of the hydraulic actuator, i.e. the measurements carried out on the test bench. During co-simulation, the MB model of the test bench and the model of the control logic and of the actuating devices run at the same time exchanging inputs and outputs values at predefined time steps.

In the following, the two co-simulating models are described.

4.3.1 MB Model of the Test Bench

The MB model of the test bench is shown Fig. 4.5. A 3D model has been considered, to account for actual contact geometry between foot and runner. Further developments in fact include the study of different solutions for the foot runner to improve reproduction of foot-ground contact.

Contact forces between foot and runner and between runner and sled have been modeled using the function CONTACT [4, 5]. Making reference to Fig. 4.6, normal contact force is calculated according to:

$$F_n = \begin{cases} k\Delta x^n + c\,(\Delta x)\,\Delta\dot{x} & x < a \\ 0 & x \geq a \end{cases} \tag{4.1}$$

Normal component of contact force is equal to zero when there isn't penetration between the bodies (i.e. there is no contact between the bodies). When bodies get in contact, force is given by an elastic and a damping contribution. The elastic component is proportional to the penetration Δx. Exponent n is called penalty factor. To avoid discontinuities, the damping coefficient c is increased from zero to its maximum value through a cubic law (Fig. 4.7a).

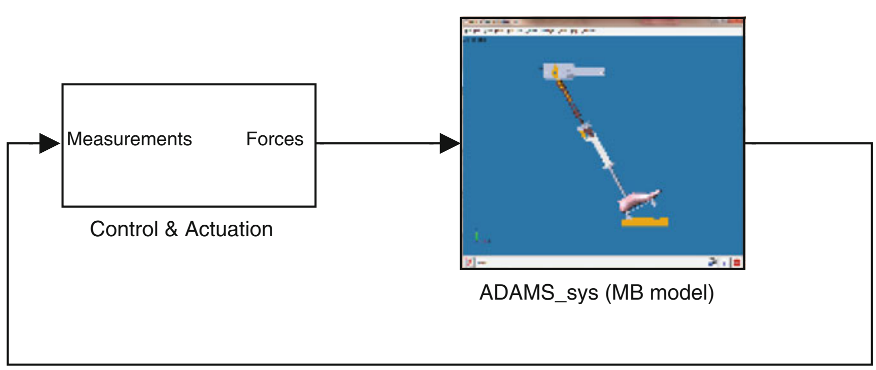


Fig. 4.4 Mechatronic model for co-simulation

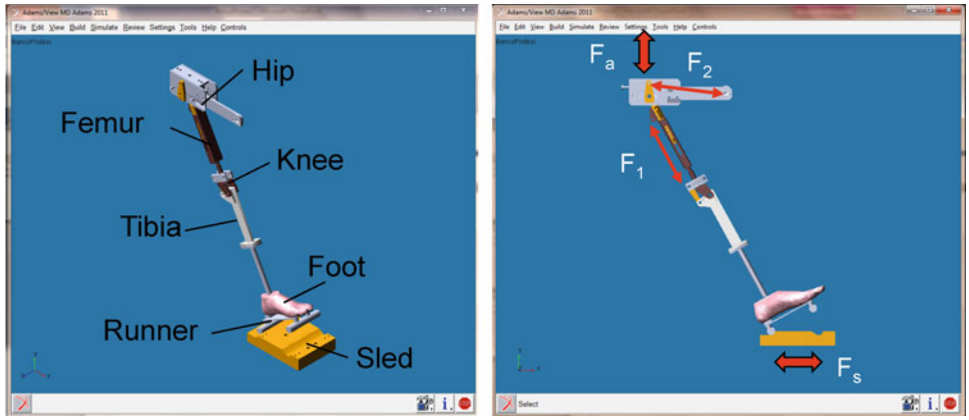


Fig. 4.5 MB model of the test bench

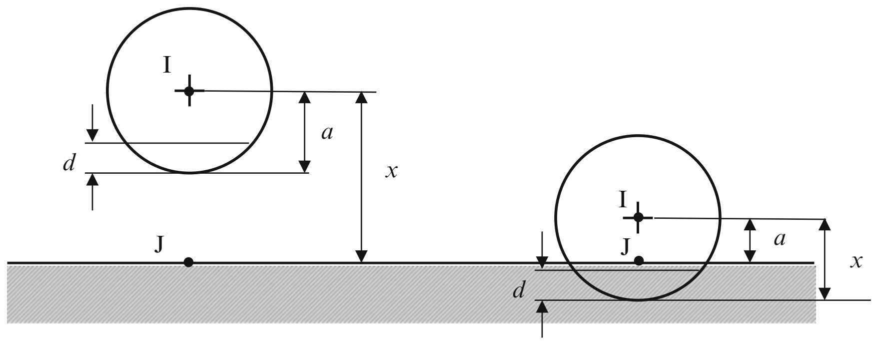


Fig. 4.6 Scheme for normal contact force calculation

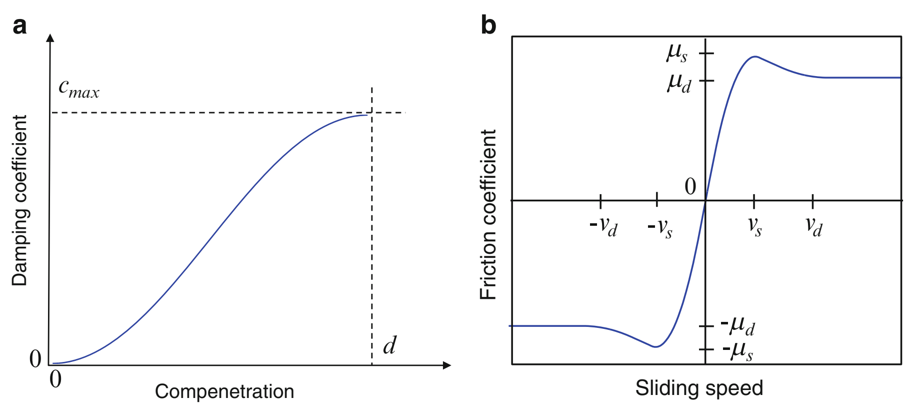


Fig. 4.7 (**a**) Damping coefficient vs. penetration; (**b**) Friction coefficient vs. sliding speed

The tangential contact force is defined according to Coulomb law:

$$F_t = \mu \left(v_{slip} \right) F_n \tag{4.2}$$

The friction coefficient is a function of the sliding speed as shown in Fig. 4.7b. Joints compliance is included into the model.

4.3.2 *Control Strategy and Model of the Actuating Devices*

4.3.2.1 Model of Actuating Devices

The hydraulic actuator (force F_a, see Fig. 4.5) driving the motion of the hip is modeled through the following equations ([6]):

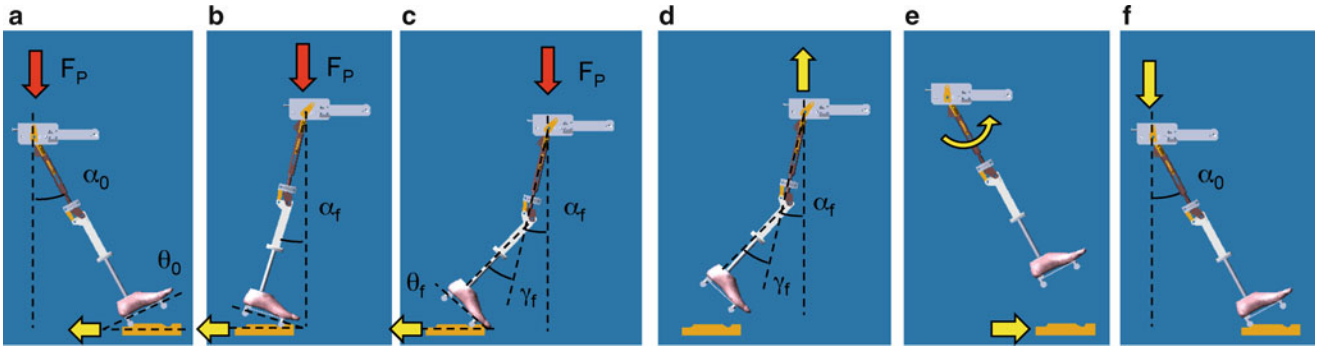


Fig. 4.8 Phases of the work cycle

$$\begin{cases} F_a = p_L A \\ \dfrac{V_0}{2\beta}\dot{p}_L + C^* p_L = K_q x_v - A\dot{y}_a \end{cases} \tag{4.3}$$

where p_L is the pressure drop across the piston rod, F_a is the force provided by the actuator, β is the bulk modulus of the oil, K_q is the servo-valve flow gain coefficient, C^* is the servo-valve total flow pressure coefficient, A is the piston area, V_0 is the volume of the hydraulic actuator chamber, x_v is the displacement of the spool valve and y_a is the displacement of the piston rod.

Dynamics of the electric motor driving the motion of the sled has instead been introduced by means of a first order time lag.

Finally, gas springs between hip and femur and between femur and tibia (forces F_1 and F_2, see Fig. 4.5) are modeled as follows:

$$F_i = \begin{cases} F_{Preload} & locked \\ k_i \Delta l_i & unlocked \end{cases} \qquad (i = 1, 2) \tag{4.4}$$

When the gas spring is locked, it provides a constant force $F_{Preload}$. On the contrary, when the gas spring is unlocked, it provides an elastic restoring force (k_i is the spring stiffness, while Δl_i is the spring deflection).

4.3.2.2 Control Logic

The control strategy has been developed assuming the following measurements:

- Force and displacement of the hydraulic actuator driving the hip motion;
- Angular speed of the electric motor driving the sled.

From this latter measurement, position of the sled is derived.

In this preliminary stage of the research, these measurements are the only available on the test bench. As a further development, sensors able to measure actual femur and tibia rotations will be added.

In the following, the phases of the developed control strategy are schematically described making reference to Fig. 4.8. As it will be explained later on, phases are triggered based on the position of the sled. Preliminary simulations have been carried out to identify the sled positions correspondent to the desired rotations of femur and tibia. At the purpose, it is worth remembering that aim of the test bench is not the complete reproduction of walking and rotations of femur and tibia are not simultaneous, but they are applied one after the other.

Phase 1 (Fig. 4.8a, b)	In the initial condition, no load is applied from the hydraulic actuator, femur is rotated of an angle $\alpha_0 = 30°$, there is no relative rotation between femur and tibia (Fig. 4.8a), sled and foot runner are in contact and the sled stands still. Then the sled is accelerated dragging the foot runner and rotating the femur clockwise. To allow femur rotation the spring between hip and femur is unlocked (force F_2, see Fig. 4.5). The phase ends when the sled reaches the position correspondent to the final rotation of femur $\alpha_f = -10°$.
	During this phase and the following, the hydraulic actuator provides a force according to ISO 22675 (see Fig. 4.9, [2]). At the purpose, the force provided by the hydraulic is controlled in feedback by means of a PID regulator acting on the servo valve position x_v. The scheme of the force feedback control is shown in Fig. 4.10.
	During this phase, there is no relative rotation of tibia with respect of femur. This is achieved by locking the gas spring placed between femur and tibia (force F_1, see Fig. 4.5).
Phase 2 (Fig. 4.8c)	Once the sled has reached the position correspondent to the femur rotation α_f, the gas spring placed between femur and tibia is unlocked to allow knee rotations. Any further rotation of femur is instead prevented by a bumpstop.
	As already mentioned, even during this phase, the hydraulic actuator provides a force according to ISO 22675.
	Phase 2 ends when the sled reaches a position correspondent to a relative rotation between tibia and femur equal to $\gamma_f = 30°$.
Phase 3 (Fig. 4.8d)	The leg is lifted and the sled is stopped. To prevent rotations of femur and tibia, both the gas springs are locked.
	During this phase, hydraulic actuator control is switched from force to displacement feedback. The scheme of displacement feedback control is shown in Fig. 4.11.
	Displacement feedback control is applied also in the next 3 phases.
Phase 4 (Fig. 4.8e)	Once the leg is completed lifted, both the gas springs are unlocked. Thus the leg swings and, after oscillations have extinguished, it gets back to the initial condition, i.e. rotation of femur equal to $\alpha_0 = 30°$ and no relative rotation between femur and tibia.
	Meanwhile the sled is accelerated to return to the initial position. Once the sled has reached its initial position, it is stopped.
Phase 5 (Fig. 4.8d)	The leg is brought down till the foot runner returns in contact with the sled.
	The gas spring placed between femur and tibia is locked.
	Then hydraulic actuator control is switched from displacement to force feedback and the work cycle re-starts from phase 1.

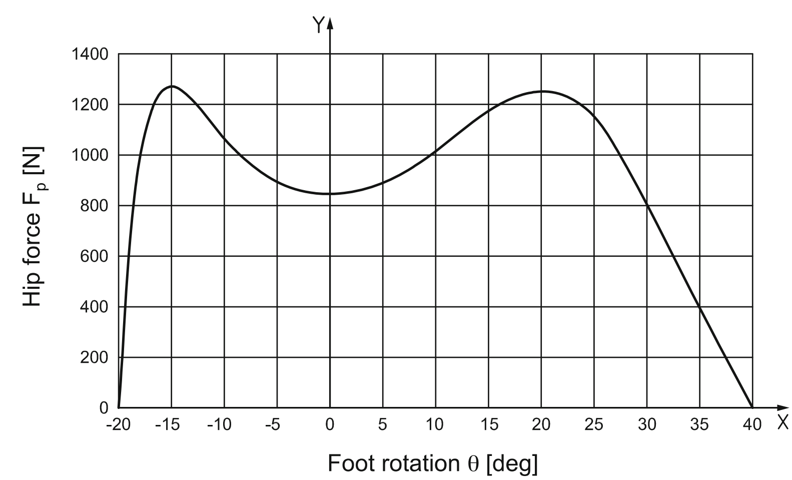


Fig. 4.9 Reference hip force vs. foot runner rotation

4.4 Simulation Results

Figure 4.12 shows the results of the implemented model during one working cycle, i.e. one step.

The hydraulic actuator force and displacement, the contact force at the foot runner-sled interface, rotations of femur and foot runner and relative rotation between femur and tibia are depicted versus the work cycle percentage. Number of phases is also reported.

Transition from force to displacement feedback control of the hydraulic actuator is clearly visible after the end of phase 2.

As it can be seen, almost half of the cycle (contrarily to real gaits) is used to lift the leg, let it swing and get it back to the initial position.

Relative rotation between femur and tibia reaches a maximum value of 30°, while the foot rotates of 75° (from −30° to 45°). These values are coherent the ones prescribed for prosthetics tests [2] and the ones collected during amputees gait analysis [3].

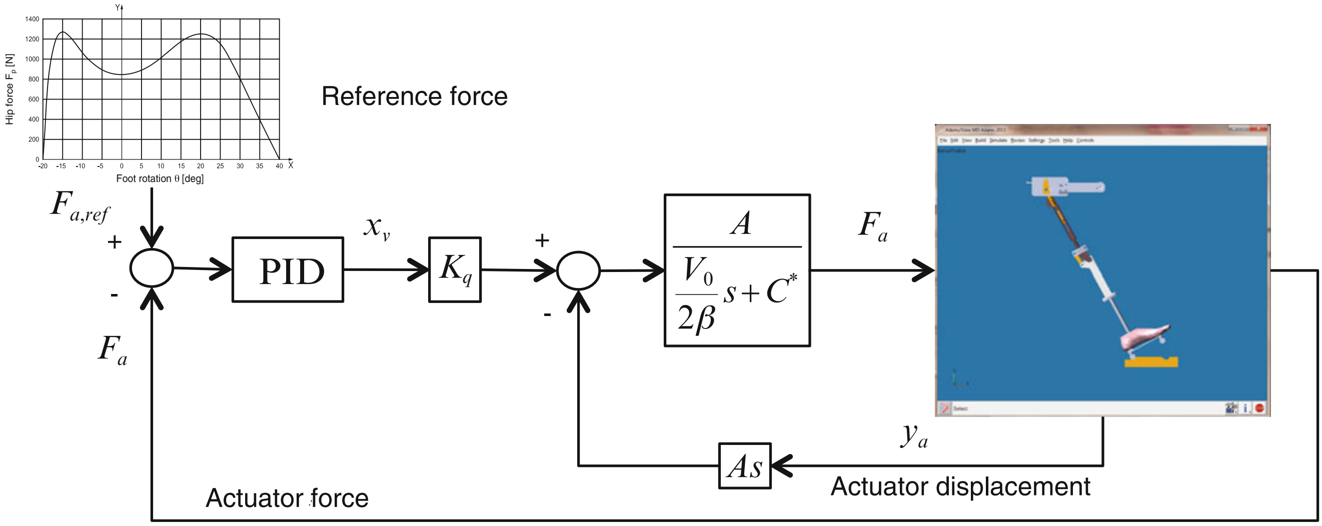


Fig. 4.10 Feedback control of hydraulic actuator force

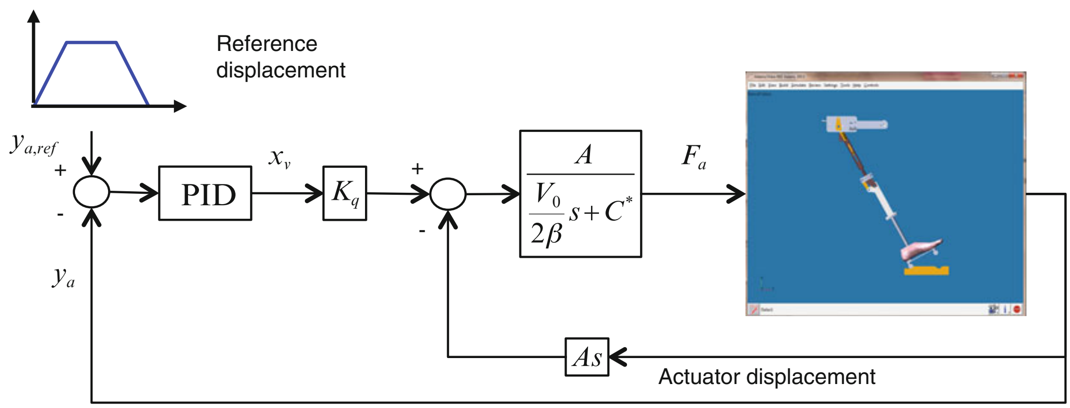


Fig. 4.11 Feedback control of hydraulic actuator displacement

Even foot-sled contact force is coherent with the one indicated in ISO 22675 [2]. Transition form heel to forefoot contact can be clearly seen in the last part of phase 1.

Actuators have been selected to allow the execution of the complete work cycle in 1 s, almost the duration of one step.

4.5 Preliminary Experimental Results

The developed model allowed to select the actuators for the test bench (electric motor, hydraulic actuator and gas springs) and to implement the control logic to driving the motion of the test bench elements. Then a prototype of the test bench has been built (Fig. 4.13).

Preliminary experimental tests have been carried out on a transtibial prosthesis. Thus, the element representing the tibia has been modified to fit the knee socket (Fig. 4.2). At the purpose, based on a 3D scan of the amputee stump, coverage for the tibia (cut in correspondence of the amputation level) has been built. The gap between coverage and the stump has been filled with polyurethane foam, to simulate the muscular tissues. For this reason, a mould for the stump was developed, as well as a system to hang the tibial structure. It must be stressed that the knee mechanism, as already mentioned, is included inside the foam coverage, which fits the knee socket. Thus when rotations between femur and tibia occur, both the foam coverage and the knee socket are bended, similarly to real walking conditions. This is of course essential for properly testing the knee socket.

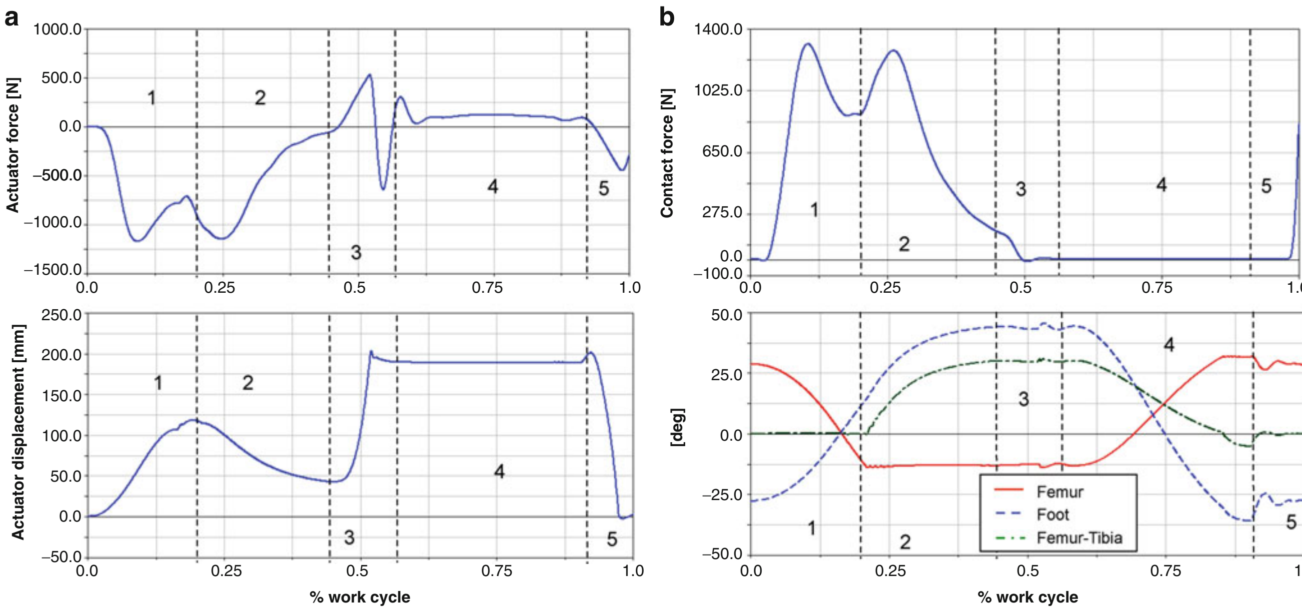


Fig. 4.12 (**a**) Hydraulic actuator motion and displacement vs. work cycle percentage; (**b**) Contact force and femur, foot and femur-tibia relative rotations vs. work cycle percentage

Fig. 4.13 Test bench

Figure 4.14 shows the different steps of design and production stages of the component. In Fig. 4.14a a 3D image of the complete structure is shown: foam has to fit the stump shape and to allow movements of the structure. Foam thickness has been chosen in order to have a higher stiffness in correspondence of the most stressed contact points: during stance phase the stump is pressed against the socket in correspondence of the knee stressing in particular the lower patellar region and the popliteal fossa, while during the swing phase, the socket is hanged to the condyle regions. In Fig. 4.14b the mould and the positioning of the structure is shown, before foam casting. Figure 4.14c shows the obtained stump, which is joined to the rest of the designed structure of the test bench (Fig. 4.14d).

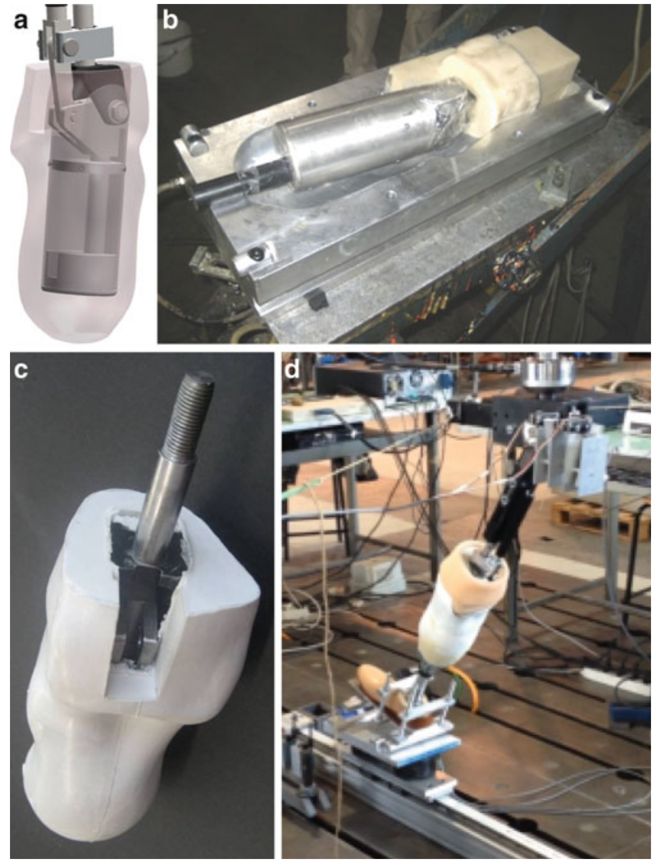


Fig. 4.14 (**a**) Design of the tibial coverage to fit the stump shape; (**b**) Placement of the tibial structure in the mould; (**c**) Obtained artificial stump; (**d**) Test bench in the final configuration

Figure 4.15 shows the results of preliminary experimental tests. A work cycle (i.e. one step) is performed, according to the control logic previously described. It is to point out that, during the experimental test, load on the hip is maintained constant, while in the previously described work cycle it was varied according to ISO 22675. Hydraulic actuator displacement and force are reported as a function work cycle percentage. The hydraulic actuator force is compared, in the lower part of the figure, with the foot-sled contact force (dashed line), obtained by summing up the measurement provided by the two load cells placed in the sled support.

As it can be seen, contact force and hydraulic actuator force are in good agreement and force feedback control is able to maintain a constant load on the hip while the sled is moved. About the 40% of the cycle is used for lifting and lowering the leg and for the swing phase. With respect of the simulated cycle the stance phase is thus longer.

4.6 Concluding Remarks

The design of a test bench for testing prostheses for legs has been presented in this paper. The test bench is constituted of a fake leg (from femur to foot). Hip motion and load are provided by a hydraulic actuator, while the foot is moved by a sled driven by an electric motor. Lockable gas springs block/allow femur rotations and femur-tibia relative rotations. In order to properly select actuators and to dimension test bench components, a model of the bench as been implemented. It is constituted of a MB model of the fake leg, a model of the actuating devices and the control logic regulating the leg motion.

After assessed the test bench performance through simulations, a prototype has been built. Some preliminary tests have been performed on a transtibial prosthesis.

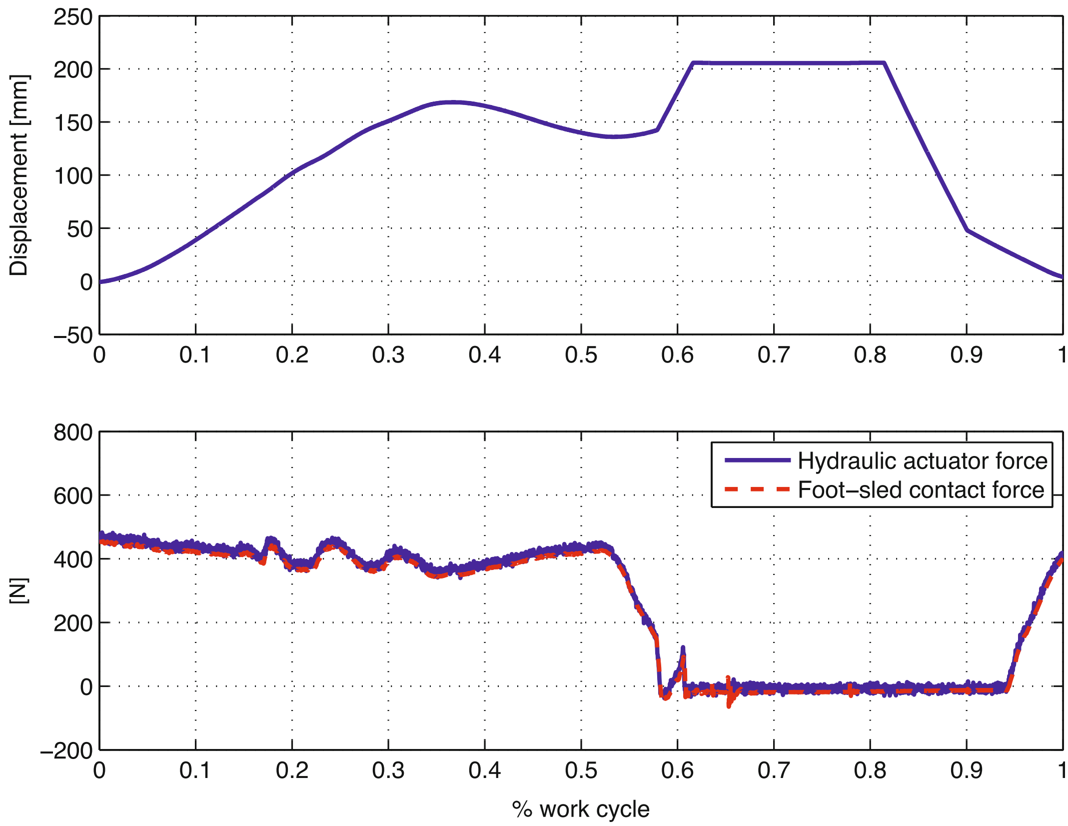


Fig. 4.15 Hydraulic actuator displacement and force and foot-sled contact force vs. work cycle percentage

References

1. ISO 10328: 2006 Prosthetics – structural testing of lower-limb prostheses – requirements and test methods
2. ISO 22675: 2006 Prosthetics – testing of ankle-foot devices and foot units – requirements and test methods
3. Perry J (1992) Gait analysis, normal and pathological function. SLACK Inc, Thorofare
4. MD Adams (2010) Online help
5. Faik S, Witteman H (2000) Modeling of impact dynamics: a literature survey. In: International ADAMS user conference, Orlando, pp 3–10
6. Merritt HE (1967) Hydraulic control system. Wiley, New York