Workshop

“ROBOTICS AND REMOTE-ASSISTANCE”

Control Applications in Rehabilitation and Medical Education

Heike Vallery, Martin Buss, M. Bernhard and T. Pröll

Institute of Automatic Control Engineering.

Technical University of Munich

Control Applications in Rehabilitation and Medical Education

Heike Vallery, Martin Buss, M. Bernhard and T. Pröll

Institute of Automatic Control Engineering.
Technical University of Munich

I) Toward a Hybrid Motor Neuroprosthesis for Gait Rehabilitation

Heike Vallery and Martin Buss

1 Introduction

In gait therapy for stroke patients, the predominant aim is restoration of mobility and independence. Therapy methods can mainly be divided in two superordinate groups:

- External guidance or supported motion, e.g. by a physiotherapist or a motor-driven orthotic device, called exoskeleton.
- Artificial induction of muscle activity to provoke motion, e.g. through Functional Electrical Stimulation (FES).

External support allows patients with strong lesions to walk in a very early stage of therapy, since it does not require the ability to stand freely. Of Exoskeletons, the HAL system [1] needs to be mentioned. The commercially available Lokomat [2] is clinically applied for gait rehabilition on a treadmill. The advanced gait trainer offers a simpler alternative [3]. It uses footplates which move the feet in a gait-like path.

Functional Electrical Stimulation (FES) has been investigated for decades (e.g. [4, 5]) and applied for neuroprostheses. It offers propioceptive feedback, which improves rehabilitation results, and a training effect for the muscles. However, severe problems arise with artificial muscle activation: Muscles are not recruited in a physiological manner, leading to substantially increased fatigue. Surface stimulation cannot offer sufficient selectivity. Furthermore, in hemiplegic patients, sensory-motor mechanisms and reflexes are modified, such that stimulation triggers unwanted responses. FES in clinical practice is predominantly applied on its own only in an advanced therapy stage, where the patient can already stand freely with a walking aid.

2 Research Idea

Motivated by investigations showing that the restoration of motor control is accelerated by a combinative therapy [6], this work investigates a cooperatively controlled combination of FES and an exoskeleton for gait rehabilitation in hemiplegic patients. This shall be called hybrid neuroprosthesis, adopting a term defined by Tomovic and Popovic, who introduced the concept of hybrid control [7, 8] and developed powered walking orthoses for paraplegic patients. They were followed by Andrews [9] and colleagues. Solomonow investigated the use of FES with unpowered orthoses for paraplegics [10].

However, the project focus here is substantially different from the previously mentioned investigations concerning paraplegia, since stroke patients on the one hand have remaining motor activity to be considered, and on the other hand the neuroprosthesis offers them not only functional walking, but also the benefit of rehabilitation.

3 Concept of the Hybrid Neuroprosthesis

3.1 Hierarchical Control Concept

The neuroprosthesis is divided into four parts:

1. Hardware: actuators (muscle stimulator and exoskeleton) and sensors.
2. Low-Level control for the coordination of the redundant actuators
3. Motion planning: The High-Level Control supervises the gait cycle and generates reference trajectories for angles and joint torques.
4. Patient interface, consisting of Input Device and Output Device.

In the following, the focus is on Low-Level control. More details can be found in [11].

3.2 Hardware: Actuators and Sensors

The exoskeleton consists of a motor-driven knee orthosis and will be supplemented by a 1-DoF hip orthosis. Sensors include angle and angular velocity sensors (goniometer-gyroscopes) [12]. Contact forces between feet and ground are measured with insoles.

3.3 Low-Level Control

The low-level control, which is depicted in fig. 1, fulfills three tasks: actuator control, actuator coordination, and integration of voluntary motor activity from the patient.

Through impedance control, the exoskeleton does not exert any forces when the patient moves his leg close to the reference trajectory calculated by the motion planning. In case of deviation, the patient feels a correction, similar to the aid of a physiotherapist.

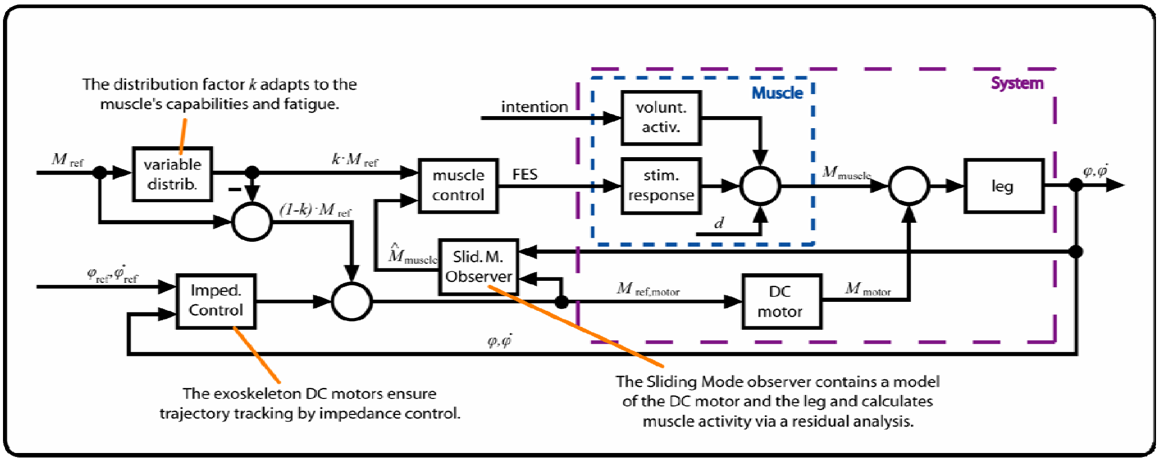
The reference torque is distributed via an adaptive convex combination, such that only a variable fraction k of the entire necessary torque is to be realized by the muscles. The rest, $(1-k)M_{\text{ref}}$, is used for a feedforward pilot control of the DC motor. A calculation method for this distribution factor k has been formulated in [13].

Coordinated with the exoskeleton, the patient’s muscles are stimulated. The muscle force depends nonlinearly on various factors such as stimulation parameters, joint angle, joint velocity, and fatigue. The predictive control approach is based on identification and simplified modeling of the muscle characteristics.

Errors in the torque trajectory calculation of, disturbances in muscle stimulation and uncoordinated patient activity can be compensated to a large extent, since the reliable DC motors of the exoskeleton are responsible for trajectory tracking.

The low-level control strategy is first evaluated in a one-dimensional experimental setup for the knee joint [13].

Future research will concentrate on integrating voluntary patient activity. In this procedure, the difference in muscle fibre recruitment between voluntary control and artificial stimulation must be taken into account.



The distribution factor k adapts to the muscle's capabilities and fatigue.

The exoskeleton DC motors ensure trajectory tracking by impedance control.

The Sliding Mode observer contains a model of the DC motor and the leg and calculates muscle activity via a residual analysis.

Figure 1: Concept for the Low-Level Control. Muscle stimulation is embedded in a torque control loop and provides a variable fraction of the necessary torque, the exoskeleton DC motors ensure trajectory tracking.

3.4 Motion Planning (High-Level Control)

A superordinate control switches between different motion patterns such as stepping or stopping and supervises the gait cycle. Angle trajectories are selected from a prefilled database. With the help of a biomechanical model of the leg, joint torques is calculated corresponding to the joint angle trajectory.

3.5 Patient Interface

The input device has to be intuitive and may not strain the patient excessively. A hemiplegic patient can generally only control half of his body and needs the unaffected arm for support on a walking aid. Thus, arm and fingers can only be of very limited use for the control of the neuroprosthesis.

It is planned to combine discrete and continuous interaction. On the one hand, the patient can command the operation mode such as standing up, stepping or stopping. This is most simply realized by switches, e.g. integrated in a crutch. On the other hand, the patient shall modify the currently executed trajectory, assisted by intention detection. A reliable possibility to deduce patient intention is to use the inclination of the trunk [14]. Such a modification of trajectories is expected to lead to an intuitive, minimally-straining interface, since it is similiar to healthy human gait, where, with the help of the so-called extrapyramidal motor system, motion patterns can be performed almost subconsciously and only need to be adjusted, e.g. to overcome obstacles.

Questions concerning the output device are: which information to transmit and which transmission to use. Essential pieces of information in healthy gait are the contact forces between foot and ground during stance. After a stroke, this information lacks on one side, which is one reason why often a disturbed balance can be observed. Therefore, these contact forces are measured, processed and made available to the patient in a simplified way via acoustic feedback. To improve intuitive learning, not only the reaction forces from the hemiplegic leg are fed back, but also those of the unaffected limb.

4 Conclusion

A concept for a hybrid neuroprosthesis for gait rehabilitation in stroke patients has been presented. Focal points in the following investigation will be the shared control of the redundant system with an integration of voluntary patient activity.

Acknowledgment

Special thanks to Dr. Quintern (Neurological Clinic Bad Aibling) as well as to Dr. Burgkart (Klinikum Rechts der Isar, Technische Universität München) for support and good ideas. The authors would also like to thank Krauth+Timmermann GmbH, Hamburg, who provide the stimulator for this work.

References

[1] H. Kawamoto and S. Kanbe, “Power Assist Method for HAL3, Estimating Operator Intention Based on Motion Information,” in *Proceedings of IEEE International Workshop on Robot and Human Interactive Communication*, Millbrae. California USA, 2003, pp. 67–72.

[2] G. Colombo, M. Jörg, and S. Jezernik, “Automatisiertes Lokomotionstraining auf dem Laufband,” *Automatisierungstechnik*, vol. 50, pp. 287–295, 2002.

[3] S. Hesse and D. Uhlenbrock, “A mechanized gait trainer for restoration of gait” *J. of Rehab. Research and Development*, vol. 37, no. 6, 2000.

[4] K. J. Hunt, “Optimal control of ankle joint moment: Toward unsupported standing in paraplegia,” *IEEE Transactions on Automatic Control*, vol. 43, no. 6, pp. 819–832, June 1998.

[5] J. Riess and J. J. Abbas, “Adaptive neural network control of cyclic movements using functional neuromuscular stimulation,” *IEEE Trans. on Rehabilitation Engineering*, vol. 8, no. 1, pp. 42–52, March 2000.

[6] K. Busch, T. Mokrusch, and D. H. W. Gr¨onemeyer, *Pilotstudie: Laufbandtherapie per funktioneller Elektrostimulatio für Schlaganfallpatienten. Benefits im Vergleich zur Physiotherapie nach Bobath,* 2004, Poster.

[7] D. Popovic and T. Sinkjær, *Control of Movement for the Physically Disabled*. Springer, 2000.

[8] D. Popovic, R. Stein, and R. Tomovic, *Nonanalytical Methods for Motor Control*. World Scientific Pub, 1995.

[9] B. Andrews and R. Baxendale, “Hybrid FES orthosis incorporating closed loop control and sensory feedback,” *Journal of Biomedical Engineering*, vol. 10, pp. 189–195, April 1988.

[10] S. Hirokawa, M. Grimm, T. Le, M. Solomonow, R. Baratta, H. Shoji, and R. D’Ambrosia, “Energy consumption of paraplegics gait using five different walking orthoses,” in *Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, vol. 3, Seattle, WA, Nov. 1989, pp. 1014 – 1015.

[11] H.Vallery and M. Buss, “Towards a Hybrid Motor Neural Prosthesis for Gait Rehabilitation,” *Journal of Automatic Control*, vol. 15 (Supplement), 2005.

[12] T. Fuhr, *Ein kooperatives, patientengeführtes Regelungssystem zur Bewegungsrestitution mit einer Neuroprothese.* VDI-Verlag, 2004.

[13] H. Vallery, T. Stützle, M. Buss, and D. Abel, “Control of a Hybrid Motor Prosthesis for the Knee Joint,” in *Proceedings IFAC World Congress, International Federation of Automatic Control*, Prague, Czech Republic, 2005.

[15] C. Azevedo and R. Héliot, “Rehabilitation of functional posture and walking:coordination of healthy and impaired limbs,” *Journal of Automatic Control*, vol. 15(Supplement), pp. 12–14, 2005.

II) Novel Applications with Repetitive Peripheral Magnetic Stimulation (RPMS)

M. Bernhard and M. Buss

1 Introduction

1.1 RPMS in rehabilitation

In central paresis, e.g. after stroke, a paresis of the arm and/or the hand reduces the quality of life dramatically. Large clinical studies showed, that 45 % of patients with completed stroke have persistent hemiparesis [1]. Cortical reorganization probably forms the basis of relearning lost motor functions and even the mature brain is capable of considerable, partly also structural modifications (reorganization, neuroplasticity) [2]. In order to activate a beneficial reorganization process in central paresis the lost or reduced proprioceptive input should be compensated. Currently physiotherapy aims to achieve such a compensation through externally applied passive movements. When the lost movements are induced by muscle stimulation, the associated proprioceptive input is much higher and corresponds closer to the lost voluntary action patterns which increases the therapeutic outcome.

A well-known method to induce movements by muscle stimulation is the functional electrical stimulation (fES) (e.g. [3]). However, the fES not only activates somatosensory nerve fibers, also cutaneous receptors are activated. This causes pain and spasticity. Therefore the usability of fES for therapeutic purposes is limited to patients with low spasticity, while fES is useful in case of paraplegia (review in [4]). Hence, in this project repetitive peripheral magnetic stimulation (*RPMS*) as a deeper penetrating, focused and painless stimulation method is used. As shown in figure 2 a stimulation coil is put on the upper arm or theforearm in order to stimulate the innervation area (terminal motor branches) of the respective muscle.

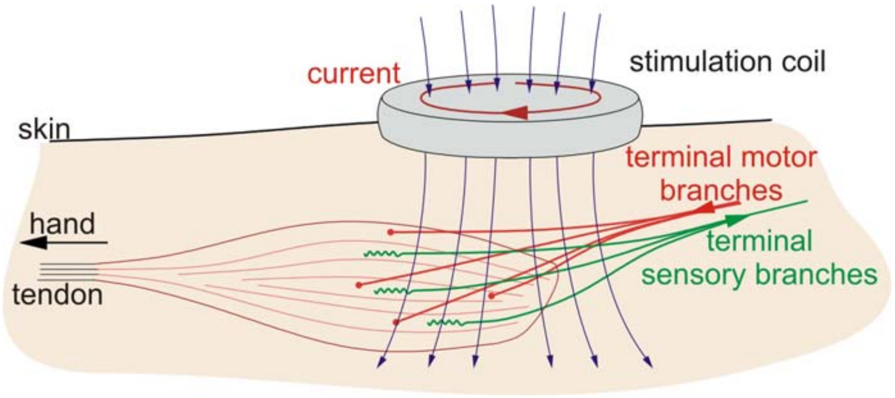


Figure 2: Principle of RPMS.

The applied field pulses are repeated at a frequency of 10-30 Hz with a maximum stimulation intensity of 100% which corresponds to a magnetic flux density of 2.0T. According to Faraday's law ($\nabla \times \vec{E} = -\partial \vec{B} / \partial t$) an electric field is induced which causes depolarization of the terminal motor branches and hence activates the respective motor units. The principle of the *RPMS* application is shown in fig. 1.

1.2 Current applications and state of the art

So far, *RPMS* is applied manually: The coil is placed on the upper arm or on the forearm, and impulses are repeated at a frequency of 20 Hz for a duration of 1.5s. Then during a break of 3s the coil is slightly displaced before the next pulse series is started. This is repeated 160 times per session. This therapy is called conditioning *RPMS*. Many neurological and clinical experimental studies have proven that the sensorimotor failures due to brain lesions can be remarkably improved by conditioning *RPMS*:

- Spasticity independent of the level of origin can be suppressed by *RPMS* [5], [6].
- Cognitive functions like spatial cognition are improved by *RPMS* [7].
- A PET-study investigating the cerebral activation shows that *RPMS* focuses the activation on a fronto-parietal circuit [8].
- A study of a postural motor performance component under relaxed state shows the *RPMS* modulates the stabilization of the elbow joint [9].

- In disturbed goal-directed motor performances, the accuracy of the performed trajectory is improved by *RPMS* [10].

Together these findings lead to the conclusion that the proprioceptive input evoked by *RPMS* modifies control systems at the spinal and cortical level. This is essential for the therapeutic effect of *RPMS*.

2 Position controlled movements of the index finger

The conditioning *RPMS* as described in paragraph 1.2 induces random movements of the forearm or the fingers. To optimize the proprioceptive inflow, and hence to improve the therapeutic outcome, it is necessary to induce coordinated and functional movements. Therefore a nonlinear closed loop position-control of the index finger as depicted in figure 3 is planned as a therapeutic *RPMS* application.

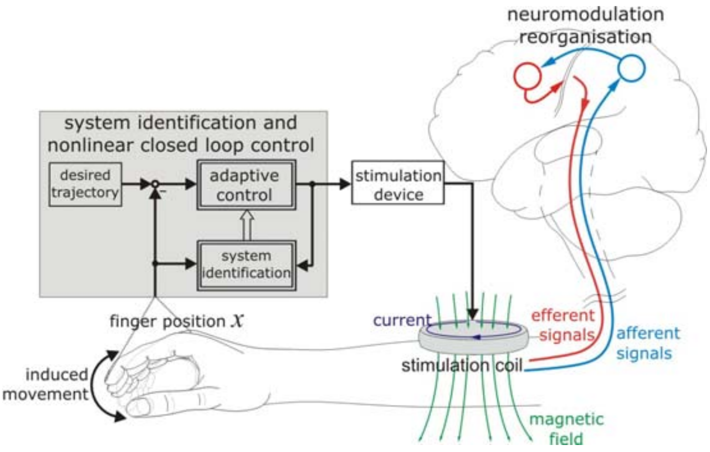


Figure 3: Nonlinear closed loop control of the index finger.

For this application the control path has to be modelled and parameterized. Since the parameters are patient specific the model must be adjustable to the respective patient. The structure as depicted in 3 seems to be appropriate to model the neuromuscular and biomechanical system.

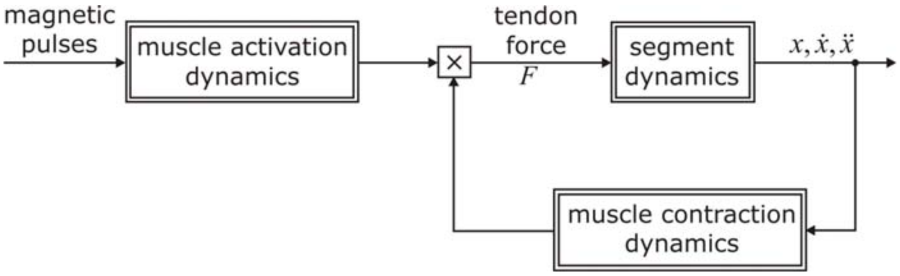


Figure 4: Appropriate structure for the model of the index finger extension

Organiza:

Patrocinan:

Colaboran:

In order to identify the muscle activation dynamics, an online identification algorithm based on the Volterra theory, the general regression neural network and the Hammerstein model as proposed in [12] will be implemented.

It is planned to extend this system in order to accomplish coordinated movements by stimulating multiple muscle groups with two or three stimulators and coils. Thus not only a finger extension but also a flexion movement could be induced.

3 Conclusion and outlook

The *RPMS* is a new painless and effective method for the rehabilitation of central paresis e.g. after stroke. It has been evaluated very positively in many clinical experimental studies. Since a lot of fundamental research on the effectiveness, the physiology and the modelling of the control path has been made, the invention and implementation of new therapy modes with *RPMS* is the logical consequence. It is assumed that the combination of *RPMS* with new methods of rehabilitation engineering will deliver good results for the treatment of sensorimotor dysfunctions after stroke.

References

[1] G. Gresham andW. Stanson, *Stroke - Pathophysiology, Diagnosis and Management*, ch. V: Stroke Therapy: Rehabilitation of the Stroke Survivor, pp. 1389–1401. New York (USA): Churchill Livingstone, New York, third ed., 1998.

[2] C. Weiller and M. Rijntjes, “Learning, plasticity and recovery in the central nervous system,” *Experimental Brain Research*, vol. 128 (1/2), pp. 134–138, 1999.

[3] H. Gollee et al., “A nonlinear approach to modelling of electrically stimulated skeletal muscle,” *IEEE Transcactions on Biomedical Engineering*, vol. 48(4), pp. 406–415, 1999.

[4] S. Agarwal et al., “Long-term user perceptions of an implanted neuro-prosthesis for exercise, standing, and transfers after spinal cord injury,” *Journal of Rehabilitaton. Research and Development*, vol. 40(3), pp. 241–252, 2003.

[5] A. Struppler, C. Jacob, P. Müller-Barna, M. Schmidt, H.-W. Lorenzen, M. Prosiegel and M. Paulig, “Eine neue methode zur frührehabilitation zentralbedingter lähmungen von arm und hand mittels magnetstimulation,” *Zeitschrift für EEG und EMG*, vol. 27, pp. 151–157, 1996.

[6] A. Struppler, P. Havel and P. Müller-Barna, “Facilitation of skilled finger movements by repetitive peripheralmagnetic stimulation (rpms) - an new approach in central paresis,” *NeuroRehabilitation*, vol. 18(1), pp. 69–82, 2003.

[7] B. Heldmann, G. Kerkhoff, A. Struppler, P. Havel and Th. Jahn, “Repetitive peripheral magnetic stimulation alleviates tactile extinction,” *NeuroReport*, vol. 11(14), pp. 3193–3198, 2000.

[8] S. Spiegel, P. Bartenstein, A. Struppler, P. Havel, A. Drzezga and M. Schwaiger, “Zentrale bewegungsverarbeitung bei spastisch-paretischen patienten nach repetitiver peripherer magnetstimulation (rpms): Eine pet-studie mit H2O-15,” *Nuklearmedizin*, vol. 39(2),p. A6, 2000.

[9] A. Struppler, B. Angerer, Ch. Gündisch and P. Havel, “Modulatory effect of repetitive peripheral magnetic stimulation (rpms) on the skeletal muscle tone (stabilization of the elbow joint) on healthy subjects,” *Experimental Brain Research*, vol. 157(1), pp. 59–66, 2004.

[10] A. Struppler, B. Angerer and P. Havel, “Facilitation of goal directed motor tasks and position sense by repetitive peripheral magnetic stimulation (rpms),” in *Proceedings of the 29th Goettingen Neurobiology Conference and the 5th Meeting of the German Neuroscience Society*, pp. 1154–1155, 2003.

[11] P. Havel, *Geregelte Induktion von Reich- und Greifbewegungen am Menschen mittels repetitiver peripherer Magnetstimuluation*. PhD thesis, Technische Univerisität München, 2002.

[12] B. Angerer, *Entwicklung von Messverfahren und nichtlinearen Identifikationsmethoden zur Anwendung an biomechanischen Systemen - Fortschritte in der Erforschung der repetitiven peripheren*

Magnetstimulation. PhD thesis, Technische Univerisität München, submitted.

III) Munich Knee Joint Simulator

T. Pröll and M. Buss

1 Introduction

1.1 Knee Joint Diagnostics

Different methods for diagnosing knee joint injuries are known. The image generating methods like CT or MRI can be used for that purpose, but they cause either harmful radiation or significant cost. Moreover, several pathologies can not be diagnosed as they do not change the shape of knee joint internals, but their properties like elastic.

The traditional, functional knee joint tests are another way to find out the pathologies of a knee joint. Basically, these tests are performed by twisting, pushing or pulling the knee joint in certain postures, so that a certain load is applied to isolated internal knee joint structures. An example for a knee joint test is the so-called varus-valgus test, where the lower leg is pushed in inwards (*valgus*, like knock-knees) and outwards (*varus*, like bow-legs) to mutually load the inner (*medial*) or outer (*lateral*) collateral ligaments.

Damaged knee joint components show a different behaviour under load than healthy components. A partly ruptured collateral ligament will hurt, thus the medical doctor can see the pathology from the patient’s reaction. If the ligament is ruptured completely, no pain occurs, but the knee joint shows an increased laxity, if it is compared with the healthy knee.

As the different patients are very different in their expression of pain, in the natural laxity of their joints and in many other parameters, no objective criterions for certain pathologies can be expressed. Moreover, many knee joint tests are more complicated and less precise than the varus-valgus test that was described above. Thus, several tests have to be applied to guarantee a profound diagnosis.

Medical doctors require many years of experience to gain enough knowledge for this kind of diagnostics. This knowledge can not be found in books or in videos, as it is a haptic knowledge, i.e. knowledge about forces and the impression of such forces. Until today, the education can only be done with phantoms - which is unrealistic - or on patients – who often refuse to serve as training objects after a few repetitions.

One way to provide haptic impressions are haptic simulators, which is also done in this project.

1.2 State of the art

Many approaches are known to simulate haptic feedback to different body parts of the user. These approaches have to be divided in the group of kinestetic and in the group of tactile displays, as they work in different manners. Tactile displays stimulate receptors in the skin and have the capability to simulate surface properties like smoothness, or temperature. Kinestetic displays apply more severe forces to the limbs of the operator. By compensating those forces, the user's muscles and tendons are loaded and stretched, which gives him/her the impression of bearing strong forces.

The commercial tactile products are mainly Braille displays that stimulate the fingertips of the user by pin arrays. Other tactile displays that are not commercially available use shear force, temperature, vibration or pressure to stimulate the fingertips [1]. All those displays can stimulate the corresponding receptors, but fail in giving a feedback that is realistic enough to be used in medical applications. For the topic of the knee joint simulator, a tactile feedback would be necessary that allows the user to find underlying bones such as the shinbone or to find the gap around the patella (kneecap), which is not possible today.

Haptic feedback to the fingertips is given by different kinds of actuated gloves, kinestetic feedback to the arms and legs can be given by different multi-DOF robots or exo-skeletons [2]. In opposite to the tactile displays, the kinestetic displays can provide realistic impressions, if the hard- and software are sophisticated enough.

2 The Knee Joint Simulator

2.1. Setup

The setup of the knee joint simulator is a hybrid one. The kinestetic feedback is given by an industrial Robot “Stäubli RX-90”. As tactile displays cannot yet provide a sufficient feedback, a passive phantom was used. This passive phantom consists of an artificial bone structure, surrounded by foam with a flesh-like consistency.

This setup allows the user to touch and palpate a virtual leg, finding relevant structures, which is a satisfactory tactile feedback. This passive shank is held by the robot, which is used to simulate the kinestetic properties of the healthy or injured knee joints.

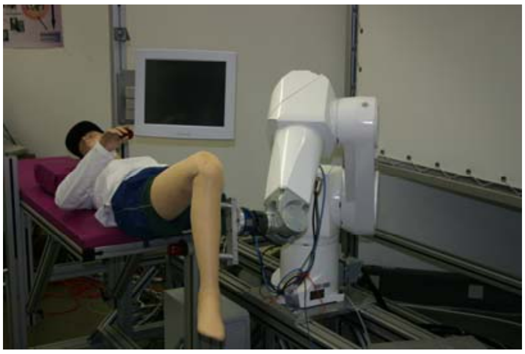


Figure 5: The setup of the knee joint simulator.

Additional to the haptic display, a graphic display and an acoustic display is implemented. The graphic display can be used to show video clips and animations of knee joint internals, whereas the acoustic display gives the user feedback about the uttering of the patient and sounds from inside the knee.

Moreover, a couch, a mannequin and an actuated thigh are implemented to give the impression of a real surgery room (figure 5).

2.2. Kinestetic Display

The kinestetic display consists of a personal computer for the control, the shank, the robot and the 6DOF force-torque sensor between them. Whenever the user applies a force to the shank, it is measured by the force-torque sensor and transmitted to the control unit. This control unit consists of a biomechanical model of the healthy and injured knee joint, so the reaction of a real knee joint to those forces can be calculated.

Afterwards, the robot is used to move the artificial shank into the calculated position. If the delays of these calculations are short enough, the user has the impression that the movement is caused directly by the applied forces.

The control and the biomechanical model are the two main challenges of the kinestetic display. The control has to ensure a free movement in flexion direction, i.e. a very low stiffness. However, in varus and valgus direction, a high stiffness has to be simulated. Neither the admittance nor the impedance control is capable to fulfil both aspects. Thus, a combination is used [3].

The biomechanical model is derived from different measurements, where the robot moved a cadaver knee, so that the correlation between the applied force and the resulting knee joint posture could be found out for all relevant DOF [4].

2.3. Graphic Display

The graphic display offers the view inside the knee joint, e.g. the relative bone movements and the soft tissue deformation of the virtual knee joint are displayed in real time (fig. 2). Therefore, the position and orientation of the biomechanical model is known, but only the orientation is used for the graphic display. As the knee joints of both displays are taken from different individuals, the direct usage of the transformation matrix would result in unrealistic gaps or interpenetrations of the knee joint components.

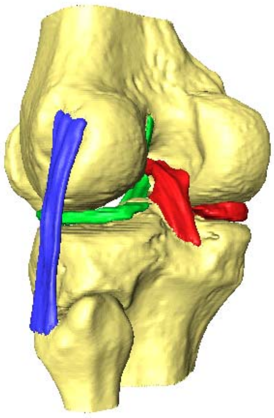


Figure 6: The view inside the knee joint

The challenge for the graphical display is to find the correct translation of the shank depending on its orientation. Moreover, the position and orientation of all other knee joint components and the deformations have to be found relative to the shank orientation. Therefore, a data acquisition process with a CT scanner had to be performed. 30 different knee joint postures were acquired and analyzed. All postures between those acquired postures can be simulated by common motion interpolation [5] and image warping algorithms [6].

References

[1] Alexander Kron, Günther Schmidt. "Multi-Fingered Tactile Feedback from Virtual and Remote Environments," haptics, p. 16, 11th Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems (HAPTICS'03), 2003.

[2] Peter Kammermeier, Alexander Kron, Jens Hoogen, Günther Schmidt. "Display of holistic haptic sensations by combined tactile and kinesthetic feedback" Presence: Teleoperators and Virtual Environments, Vol. 13, p. 1-16, 2004

[3] Martin Frey, Jens Hoogen, Rainer Burgkart, Robert Riener. "Physical interaction with a virtual knee joint – The 9 DOF haptic display of the Munich Knee Joint Simulator. Presence, 2004

[4] Martin Frey, Rainer Burgkart, Felix Regenfelder, Robert Riener. "A new robot based setup for exploring the stiffness of anatomical structures" Proceedings of the International Society of Biomechanics Congress (ISCSB), p. 111, 2003

[5] Charles Rose, Michael Cohen, Bobby Bodenheimer. "Verbs and Adverbs: Multidimensional Motion Interpolation". IEEE Computer Graphics and Applications, Vol. 18, p. 32-41, 1998

[6] F.L. Bookstein. "Principal warps: Thin-plate splines and the decomposition of deformations". IEEE Trans. Pattern Anal. Mach. Intell., Vol. 11, p. 567-585, 1989