1 Abstract

To improve arthroscopic skills, the orthopaedic residents train on medical simulators and cadaveric training courses. Some medical simulators provide feedback while cadaveric specimen are most realistic training environments. Providing objective feedback while training on cadaveric specimen, will be an major improvement to the cadaveric training courses. The goal of this project is to develop a system that allows objective force feedback during arthroscopic skills training on cadaveric specimen. Therefore an analysis of the procedure is made, a theoretical model is set up and from multiple concepts a 6DOF force sensor method is chosen. We developed a 6DOF force platform “Force Measurement Table”(FMT), which measures forces and torques in the range of 0-750N, with an accuracy of ±0.32N. The FMT was calibrated, and shows well able to distinguish two statically provided independent forces. A pilot study with two intermediates shows a significant difference between the average of forces exerted by the arthroscope (2.4 ±1.2N) and the instrument with arthroscope (4.5 ±2.4N).Indicating the deviations of force during navigation rounds are mostly exerted to the specimen by the probing instrument. The technology of the FMT, shows high potential for the use of force measurements during cadaveric training courses. For exact dynamical measurements, a accurate tracking system needs to added to the FMT.

2 Introduction

New operative surgical techniques are developed continuously. The main focus of these innovations is the optimal care for the patient. An example is arthroscopic surgery which decreases postoperative pain and increases patient recovery. A drawback is that these innovations demand significantly more surgical skills of the surgeon because of: the reduced visibility, reduced degrees of freedom of instruments, different eye-hand coordination, loss of force feedback and loss of tactile feedback. Learning to master these skills demands many hours of training [1].

Traditionally, surgical skills are trained in the regular clinical environment, the so-called apprenticeship model [2]. These training methods have been proven to be insufficiently adjusted to the new surgical techniques. The apprenticeship model is recognized as not the most efficient and safe, because training takes place on patients. This compromises patient safety, especially during the first procedures performed by the residents. [2]

New methods are introduced to train arthroscopic skills away from the operating room, such as cadaveric training courses and training on medical skills simulators [1, 3, 4]. Training skills outside the operating room can increase the skills of surgical residents, and also decreases the risk of medical errors. This way the patient safety is more secured from first-time mistakes. Also the resident is allowed to make errors without adverse consequences, which provided that the right feedback in training performance is given, lead to steeper learning curves and thus a more efficient training curriculum.

Cadaveric training courses proved a safe and realistic training environment. A drawback is that feedback on perfor-
mance and assessment relies on the subjective rating of the supervising professional. Also immediate feedback during an exercise is usually not possible, which makes it difficult to pinpoint the exact cause of tissue damage as feedback to the trainee.

Another class of training means are medical simulators, which seems a promising modality. The advantage is that these devices can provide direct objective feedback during an exercise and objective performance assessment to the trainee. Similarly to cadaveric material, they offer a safe and efficient training environment. A drawback is that simulation of real-life human tissue, especially the interaction between human tissue and cutting instruments is a challenge in both phantom models and virtual reality environments. More complex procedures, such as anterior cruciate ligament reconstruction and prosthesis placements, are mostly trained on cadaveric tissue because simulators do not have a realistic resemblance of the procedure.

Concluding, an ideal situation could be created by combining the advantages of the simulator environments and cadaveric material. This will provide a safe and realistic environment and results in optimal training efficiency by providing direct objective feedback and objective performance assessment to the surgical trainee.

The goal of this master thesis is to design and evaluate a system that allows objective feedback during arthroscopic skills training on cadaveric material.

3 Methods

The design could be split into two elements: the design of a fixation mechanism that allows any cadaveric specimen to be fixated to the environment, and the design of the sensory equipment which allows the system to give direct feedback to the user during an exercise.

3.1 Requirements

All requirements are presented and discussed, which are summarized in Table 1.

3.1.1 Fixation

Relevant bone characteristics for proper fixation of a wide range of cadaveric specimens is summarized in Table 6 (Appendix A). Most bones are cylindrically shaped, but for the hip and shoulder a flat bone is used for the fixation. The fixation system needs to withstand all forces applied during surgery.

Maximum load

This implies resisting a maximum torque moment that consists of the maximum force applied by the trainee multiplied with the maximum distance between force and fixation system. The maximum load on the force measurement system will predominantly depend on the joint stress force and the mass of the specimen. The maximum mass of a specimen is 12kg [13] and the maximum joint stress force 142N [16] at a maximum distance of 0.7m from the attachment of the femur bone towards the system (Table 7, Appendix A). We can calculate the maximum load as followed:

\[ M_{\text{max, tot}} = F_{\text{joint stress}} \cdot r_{\text{leg}} = 142 \cdot 0.7 = 99.4 \text{Nm} \]  \hfill (1)

\[ F_{\text{max, tot}} = F_{g, \text{spec}} + F_{\text{joint stress}} = 262 \text{N} \]  \hfill (2)

3.1.2 Providing feedback

The goal of providing feedback to trainees is to give the trainee objective information on their performance during skills training [5]. We can distinguish two types of feedback. Direct feedback that is provided directly after the trainee makes an error by signalling a warning to the trainee. Indirect feedback, also referred to as objective assessment, that is provided directly after the trainee has ended his/her exercise by objective parameters. These parameters are determined from continuous measurements of the entire exercise. Time and motion parameters are most frequently applied [6, 7]. Besides these, tissue handling which involves forces exerted to the tissue seems to provide important information about the trainees performance during arthroscopic surgery [8, 9].

As several systems are available for motion and time tracking that can be applied also in cadaver training, the focus of this thesis is on the implementation of direct force feedback during arthroscopic training. This is not a straightforward task, as one of the goals is to keep the system very affordable and time efficient in operation.

3.1.3 Direct feedback

For direct force feedback, the forces applied by the surgeon on delicate tissue is the most important parameter. Therefore, it is necessary to have measurement data indicating the value and location of the applied force. The data should be able to distinguish the force exerted by the tip of the instrument and the force exerted by the arthroscope from the other forces applied to the specimen (Appendix A). During arthroscopy we can distinguish four types of forces that are applied to the specimen. The gravitational forces due to the mass of the specimen, the joint stressing forces which are applied to the specimen to increase the operative joint space, the portal forces exerted by the tissue on the instrument at place of joint entrance through the skin, and the instrument tip forces on to manipulate the tissue. The most interesting force for direct force feedback is the instrument tissue force. The orthopaedic trainee receives direct feedback on how to gingerly manipulate the tissue with the right amount of force. There is little knowledge on precise tissue handling forces during arthroscopy, according to Tuijthof [10] the forces used to probe the menisci is between 1.3 and 5.5N. This can be used as an indication of all the tissue handle forces.
Table 1. Table showing all requirements for the system.

<table>
<thead>
<tr>
<th>Requirement</th>
<th>Quantification</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shape</td>
<td>Flat and cylindrical</td>
<td>≤ 30mm</td>
</tr>
<tr>
<td>Maximal Force</td>
<td>$F_{\text{max, tot}}$</td>
<td>≤ 262N</td>
</tr>
<tr>
<td>Maximal Torque</td>
<td>$M_{\text{max, tot}}$</td>
<td>≤ 99.4N</td>
</tr>
<tr>
<td>Direct feedback</td>
<td>Tissue forces</td>
<td>0-19.5N</td>
</tr>
<tr>
<td>Accuracy</td>
<td>minimal error</td>
<td>≤ 0.5N</td>
</tr>
<tr>
<td>Speed</td>
<td>Reaction time</td>
<td>0.25s</td>
</tr>
<tr>
<td>Cost</td>
<td>Overall fabrication</td>
<td>Cheapest system which full fills all other design criteria</td>
</tr>
</tbody>
</table>

3.1.4 Range and Accuracy

According to Tuijthof [10] the forces when probing menisci is between 1.3N and 5.5N. And according to Obeid [11] the cartilage strength is maximum 19.5N/mm². The instrument contact area is around 1mm², we can expect a maximum contact force allowed on the cartilage around 19.5N. We can assume that the tissue forces allowed to be measured is between 0-19.5N. Than a measurement accuracy of ±0.5N is enough for an accurate force measurement.

3.1.5 Processing speed

Processing speed is the time we allow between the error made by the trainee and the signal given to the trainee. We will set this at the reaction time of the user (0.25s) this should be enough for the user to recognise which action causing which error.

3.1.6 Cost

In the design, the total cost of the system needs to be taken in consideration, including the sensors and fabrication. The cheapest system that does full fill the other requirements will be chosen.

3.2 Conceptual design

3.2.1 Fixation

Fixation of the cadaver specimen can be performed in two ways: shape enclosed fixation and friction fixation. With shape enclosed fixation the same amount of force is applied on the complete clamped surface. Hereby the shape of the clamping system has to be variable to adjust to the shape of the bone. In the case of friction fixation, the clamping system has one predesigned shape and the clamping force needs to be adjusted to be high enough to withstand the worst case loading.

3.2.2 Analysis surgical procedure

Analysing the arthroscopic procedures commonly performed, we can distinguish two main surgical actions. The first action is navigation. The surgeon navigates a probe and an arthroscope throughout the joint usually in a predetermined trajectory with the aim to confirm the diagnosis (Fig. 1) [1]. The skills of triangulation and portal placement are the most important skills to be learned to perform this action adequately [12]. The second action to be mastered is tissue removal. This implies cartilage or soft tissue is to be cut with a shaver or a punch. To work out a more detailed conceptual design, the focus will be on arthroscopic knee procedures because this is the first arthroscopic procedure to be trained in the curriculum of an orthopaedic resident.

3.2.3 Theoretical Model

To evaluate the best design for measuring forces during arthroscopic procedures on cadaveric specimen, it is necessary to know the interaction between forces in the system consisting of an instrument, arthroscope and specimen. A theoretical model will be described in the following paragraph showing a theoretical relation between the forces on the complete system.

In Fig. 2 a sketch of the system is shown. The system consist of the specimen having global coordinate system $(x_s, y_s, z_s)$, an instrument having a local coordinate system $(x_i, y_i, z_i)$, and an arthroscope having a local coordinate system $(x_c, y_c, z_c)$. Axis $z_i$ coincides with the longitudinal axis of the instrument, and $x_i$ and $y_i$ axes are perpendicular to the instrument axis such that the $y_i, z_i$-plane coincides with the gravita-
3.2.4 Analytical model

Using the precondition of static equilibrium, we can derive the following Newton-Euler equations.

\[ \sum F_x^i = F_{\text{hand},x}^i + F_{\text{tip},x}^i - F_{\text{port},x}^i = 0 \]  
\[ \sum F_y^i = F_{\text{hand},y}^i + F_{\text{tip},y}^i - F_{\text{port},y}^i - F_{\text{g,inst}} \cdot \cos \theta = 0 \]  
\[ \sum F_z^i = F_{\text{hand},z}^i - F_{\text{tip},z}^i + F_{\text{g,inst}} \cdot \sin \theta = 0 \]

We can rewrite the Eqs (3 - 9) as portal and tip force as function of the moments and forces on the handle:

\[
\begin{pmatrix}
\frac{1}{b} \cdot (F_{\text{hand},y}^i \cdot a + M_{\text{hand}}^i) \\
\frac{1}{b} \cdot (F_{\text{hand},x}^i \cdot a + M_{\text{hand}}^i) \\
\frac{1}{b} \cdot (F_{\text{hand},x}^i \cdot a + M_{\text{hand}}^i) + F_{\text{g}} \cdot (a + 2b - c) \cdot \cos \theta \\
F_{\text{hand},z}^i - F_{\text{g}} \cdot \sin \theta
\end{pmatrix}
\]
The force on the tip of the instrument is the force of interest, because this is the force that could result in tissue damage. The measurement system needs to measure or to calculate from indirectly measured forces. Four concept scenarios are chosen to analyse the optimal sensor placement.

### 3.2.5 Tip force sensors

In the first scenario, a force sensor is placed on the tip of both instruments. The advantage of measuring the forces on the tip is that there is no analytical step needed after measurement, the measurement data is the force on the tip. But a disadvantage is that the instruments needed to be adjusted to be able to measure forces. This could change the perception of the trainee. Another disadvantage is that knowledge of other forces and position cannot be gained from the measurement data. From Eq.(10) and Eq.(11), the relation between measured force and forces in the model can be described as followed:

\[
\begin{pmatrix}
F_{\text{tip},x}
F_{\text{tip},y}
F_{\text{tip},z}
\end{pmatrix}
= \\
\frac{1}{b} \begin{pmatrix}
F_{\text{hand},x} \cdot (a + b) + M_{\text{hand},x}^i \\
F_{\text{hand},y} \cdot (a + b) + M_{\text{hand},y}^i + F_g \cdot (a + b - c) \cdot \cos \theta \\
0
\end{pmatrix}
\] (11)

The main advantage is that we do not need extra sensors to distinguish between tip and portal forces. But the measured forces also need some calculations and assumption with the theoretical model. And a sensor needed to placed on the instrument, which could change the perception of the trainee and makes it difficult to interchange between instruments. A big disadvantage is that the sensor needed to be very small, and probably be very expensive.

### 3.2.7 3DOF bone sensor

By implementing a force sensor outside of the training environment, for instance between the clamping system and ground of the hole system, the tissue forces on the tip of the instrument can be analysed form the 3D data. The sensor should measure forces in the x,y,z-direction of the global coordinate system. By analysing the data from the sensor together form data of a tracking system, the tissue forces on the tip of the instrument can be calculated. The measured force data \(F_{\text{measure}}\) in the 3DOF force sensor is the combination of the portal forces and the forces on the tip of the instrument and scope. We can describe them as followed:

\[
F_{\text{measure}} = \begin{pmatrix}
F_{\text{port},x}^i + F_{\text{tip},x}^i \\
F_{\text{port},y}^i + F_{\text{tip},y}^i \\
F_{\text{port},z}^i + F_{\text{tip},z}^i
\end{pmatrix} \cdot R^s + \begin{pmatrix}
F_{\text{port},x}^c + F_{\text{tip},x}^c \\
F_{\text{port},y}^c + F_{\text{tip},y}^c \\
F_{\text{port},z}^c + F_{\text{tip},z}^c
\end{pmatrix} \cdot R^s (14)
\]

Whereby \(R^s\) and \(R^c\) is the rotational matrix converting the force in the local coordinate system to the global coordinate system. The rotational matrix can be calculated with the position and orientation of the instruments (Appendix J). So measurement data of the position and orientation of the instrument is needed.

\[
\begin{align*}
R_x &= \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos \phi & -\sin \phi \\ 0 & \sin \phi & \cos \phi \end{pmatrix} \\
R_y &= \begin{pmatrix} \cos \theta & 0 & \sin \theta \\ 0 & 1 & 0 \\ -\sin \theta & 0 & \cos \theta \end{pmatrix} \\
R_z &= \begin{pmatrix} \cos \psi & -\sin \psi & 0 \\ \sin \psi & \cos \psi & 0 \\ 0 & 0 & 1 \end{pmatrix}
\end{align*}
\] (15)

\[
R^s = R_x \cdot R_y \cdot R_z
\] (16)

So if we want to distinguish \(F_{\text{tip},x}^i\) and \(F_{\text{tip},x}^c\) more measurements or assumptions are needed in this scenario. First,
we need to be able to distinguish between the force of the scope and the force of the instrument. We could assume that the scope, moves significant less than the instrument during tissue manipulation. If we look at a basic navigational round during knee surgery, the surgeon orientates the scope before a specific tissue manipulating task, in the optimal position. This is the position where he has the best view on the area where the task is performed. During this tissue manipulating task, we can assume he hold the scope still at the optimal position and moves with the scope. If this assumption is true, the deviation of forces can be assumed to be caused by the instrument. During a specific task:

$$\Delta F_{\text{measure}}^s = (F_{\text{port},i}^s + F_{\text{tip},i}^s) \cdot R^s$$  (17)

In Appendix B an analysis is made for the instrument movement on the data gathered with the PASSPORT arthroscopic knee simulator [1]. The analysis show that the arthroscope moves significant less during some parts of the navigational round. The forces on the tip of the scope can be assumed to be significantly lower than the forces on the tip of the instrument, because the scope is not been used to manipulate the joint tissue. The tip of the scope is assumed move freely in open space. If this assumption is true the measured force is only dependent on the force on the tip of the instrument and portal of the instrument (Eq.17).

Second, we need to be able to distinguish between the force caused by the portal and the forces caused by the tip of the instrument. Because the deviations of the measured forces during a specific task are depending on the force on the portal and the force on the tip. The force on the tip can be modelled as a spring system, with the assumption that the portal force is dependent on the displacement of the portal.

$$F_{\text{port},i}(u) = k \cdot u$$  (18)

Whereby $u$ is the displacement of the portal, and $k$ is the spring constant, in case of the skin of the portal can be assumed to be modelled as a linear spring.

From the data of the 3DOF bone sensor, we are able to analyse the tip forces of the instrument. Provided that, with an extra measurement the forces applied to the portal is measured. And provided that one of the instruments applies significant less force on the cadaveric specimen. Advantage of the 3DOF force sensor is that a lot of forces can be gathered form the force data of the sensor. Also the instrumentation and the cadaveric specimen do not have to be adjusted for the force measurement. But a disadvantage is that the analytical approach relies on assumptions. And a new extra sensor is needed to measure the forces applied by the instrumentation on the portals.

$$F^s_{\text{measure}} = (F^s_{\text{port},i} + F^s_{\text{tip},i}) \cdot R^s$$  (20)

$$M^s_{\text{measure}} = (F^i_{\text{port},i} \cdot R^{cs}) \times r^s_{\text{port},i} + (F^i_{\text{tip},i} \cdot R^{cs}) \times r^s_{\text{tip},i}$$  (21)

Now we can rewrite the measured force and torque vector to one local coordinate system, the coordinate system of the instrument. This is done with the inverse rotational matrix $R^{si}$.

$$F^i_{\text{measure}} = F^s_{\text{measure}} \cdot R^{si} = F^i_{\text{port},i} + F^i_{\text{tip},i}$$  (22)

$$M^i_{\text{measure}} = M^s_{\text{measure}} \cdot R^{si} = F^i_{\text{port},i} \times r^i_{\text{port},i} + F^i_{\text{tip},i} \times r^i_{\text{tip},i}$$  (23)
Since \( b \) is the overall distance between portal and tip of the instrument, we can describe vector \( \vec{b} \) as the distance vector between portal and tip in the global coordinate system.

\[
\vec{b} = r_{\text{port},i} - r_{\text{tip},i} \tag{24}
\]

Then substitution Eqs.24 into Eqs.23:

\[
M_{\text{measure}} = \begin{bmatrix} F_{\text{port},x}^i \times r_{\text{port},i}^j + F_{\text{tip},x}^i \times (r_{\text{port},i}^j - \vec{b}) \\
F_{\text{port},y}^i \times r_{\text{port},i}^j + F_{\text{tip},y}^i \times (r_{\text{port},i}^j - \vec{b}) \\
F_{\text{port},z}^i \times r_{\text{port},i}^j + F_{\text{tip},z}^i \times (r_{\text{port},i}^j - \vec{b}) \end{bmatrix} = \begin{bmatrix} (F_{\text{port},x}^i + F_{\text{tip},x}^i) \times r_{\text{port},i}^j - F_{\text{tip},x}^i \times \vec{b} \\
(F_{\text{port},y}^i + F_{\text{tip},y}^i) \times r_{\text{port},i}^j - F_{\text{tip},y}^i \times \vec{b} \\
(F_{\text{port},z}^i + F_{\text{tip},z}^i) \times r_{\text{port},i}^j - F_{\text{tip},z}^i \times \vec{b} \end{bmatrix} \tag{25}
\]

Since Eqs.22:

\[
M_{\text{measure}} = F_{\text{measure}}^i \times r_{\text{port},i}^j - F_{\text{tip},i}^j \times \vec{b} \tag{28}
\]

Since the vector \( \vec{b} \) in local coordinates can be described as:

\[
\vec{b} = \begin{bmatrix} 0 \\ 0 \\ b \end{bmatrix} \tag{29}
\]

Eqs.28 can then be rewritten as:

\[
\begin{bmatrix} b \cdot F_{\text{tip},y}^i \\ b \cdot F_{\text{tip},x}^i \\ 0 \end{bmatrix} = \begin{bmatrix} M_{\text{measure},x}^i \\ M_{\text{measure},y}^i \\ M_{\text{measure},z}^i \end{bmatrix} + \frac{1}{3} \left( \begin{bmatrix} F_{\text{measure},y}^i \cdot r_{\text{port},x}^j - F_{\text{measure},z}^i \cdot r_{\text{port},y}^j \\ F_{\text{measure},z}^i \cdot r_{\text{port},x}^j - F_{\text{measure},x}^i \cdot r_{\text{port},y}^j \\ F_{\text{measure},x}^i \cdot r_{\text{port},y}^j - F_{\text{measure},y}^i \cdot r_{\text{port},x}^j \end{bmatrix} \right) \tag{30}
\]

The torque along the axis of the instrument is assumed to be zero in the theoretical model, and \( F_{\text{port},z}^i = 0 \). We can describe the tip force on the instrument as followed:

\[
\begin{bmatrix} F_{\text{tip},x}^i \\ F_{\text{tip},y}^i \\ F_{\text{tip},z}^i \end{bmatrix} = \begin{bmatrix} \frac{1}{3} \left( M_{\text{measure},x}^i \cdot r_{\text{port},y}^j - M_{\text{measure},z}^i \cdot r_{\text{port},y}^j \right) \\ \frac{1}{3} \left( M_{\text{measure},y}^i \cdot r_{\text{port},x}^j - M_{\text{measure},x}^i \cdot r_{\text{port},x}^j \right) \\ F_{\text{measure},z}^i \end{bmatrix} \tag{31}
\]

And:

\[
\begin{bmatrix} F_{\text{port},x}^i \\ F_{\text{port},y}^i \\ F_{\text{port},z}^i \end{bmatrix} = \begin{bmatrix} F_{\text{measure},x}^i - F_{\text{tip},x}^i \\ F_{\text{measure},y}^i - F_{\text{tip},y}^i \\ 0 \end{bmatrix} \tag{32}
\]

The advantage of using the 6DOF bone force sensor is that it can distinguish between portal and tip forces with some calculations from the theoretical model. Also we can measure all forces and locate all forces and directions including the joint stress force. And we do not need an extra sensor or make adjustments to the instrumentation. Although the 6DOF force sensor approach also relies on calculations and assumptions in the theoretical model, it is less than the 3DOF bone sensor. And the assumption can be tested in an experimental setting.

### 3.2.9 Chosen conceptual design

Considering the fact that physically modifying the instruments could change the perception of the surgeon, direct measurement at the tip and measurement in the handle are only feasible if modifications are very small. However, this would imply that only modified instruments could be used to measure the forces. This would not contribute to an affordable set-up and does not allow other unmodified instruments to be used in the system. The 6DOF force sensor and 3DOF force sensor are the only conceptual designs that does not change the instrument itself. For the 6 DOF bone sensor, we do not need more sensors to distinguish between tip (\( F_{\text{tip}} \)) portal force (\( F_{\text{port}} \)) and it allows to calculate the amount of force, the position of the force and the direction of the force vector (Eqs.31,32) . Also it has the ability to measure joint stressing forces. The 3DOF bone sensor does need a extra sensor to distinguish between portal and tip forces and which relies on the assumption that the skin can modelled as a linear spring. By these considerations the 6DOF bone force sensor is chosen to be the best method to measure and distinguish the portal an tip forces during cadaveric training of arthroscopic surgery. Three assumptions are critical for the design of the 6DOF sensor. First, we can decouple forces and moments in three directions. Second, we can distinguish between tip force and portal force. And third, the forces by the arthroscope are significant less than the forces on the instrument.

### 3.3 Prototype

The principle of the 6DOF bone force sensor is that the system is able to decouple the torques and forces in three directions. Therefore a prototype was made which uses a bending beams, which is less stiff in one direction as in the other directions. By measuring the strains distance bending beams due to the force exerted to the system, an accurate measurement of the force can be done. The prototype consists of two main designs: The force measurement table (FMT) and the clamping system. First the mechanical construction will be
explained, than the chosen sensors measuring the strain distance will be given.

For mathematical calculations and data acquisition and processing of measurements, MATLAB® version 8.0.0.783 (R2012b) (The Mathworks, Natick, USA) was used. For the dimensional design and finite element calculations SolidWorks® Premium 2011 X64 Edition, (Dassault Systèmes S.A., France) was used. In Appendix K all technical drawings of all part of the FMT are given.

3.3.1 Force measurement table

The main dimensions of the FMT are 312mm x 260mm x 40mm, and the supporting frame has a height of 170mm. The FMT consist of three square frames connected together with four bending beams (Fig.4). The frames are water-cut from solid lightweight aluminium plates. The use of bending beam was chosen over the use of other systems, because it has less influence of hysteresis, no friction and the system can be made compact. The aluminium bending beams were given dimensions such that they are compliant in one direction and stiff in the other directions, and also stiff to external torques. In Appendix C the calculations and dimensions on the bending beams are given. Each square frames allows stretching in a particular direction. By measuring the strain between all three squares, and a priori knowledge of the stiffness of FMT, the applied forces can be calculated (Fig.5).

Fig. 4. Pictures of the force measurement table solid works model. In the top picture the x,y,z-coordinate system of the FMT is drawn.

Withstanding forces an moments

With the FEM simulation toolbox in SolidWorks the stiffness and the strength of the measurement table was calculated for worst case scenarios using the expected maximum forces and moments as calculated in Eq. (1) and Eq. (2). With the Solidworks model (Fig.6), the factor of safety (FOS) is calculated from the von Mises graph, which is 1.2 in this worst case scenario the factor. Thus the chosen dimensions in combination with the material allow the resistance of the maximum loads while staying in the linear-elastic zone of the aluminium material. The construction of the frames is such that the maximum displacement between the square rings is limited to be 3 mm at which the FOS is 1. So the system does not allow to reach above the maximum strength. The stiffness of the force measurement table is calculated in Solidworks with the simulation toolbox: kx=250N/mm, ky=214N/mm and kz=248N/mm. Using the maximum 3mm distance of each bending beam, the system can measure forces up to approximately 750N.

Force measurement

A couple of methods to measure force were investigated. The three sensors are shown in Table 2. The hall effect
sensor was chosen to measure the forces because it is relatively cheap and is within the range of accuracy needed. Also it is possible to optimize the stiffness of the system. By using bending beams, where the stiffness in one direction is less than the other directions, the change in distance due to the force extended to the beam can be measured by the hall effect sensor. The use of bending beam was chosen over the use of other systems, because it has less influence of hysteresis, no friction and the system can be made compact. Since the accuracy of the hall effect on 3mm distance of the sensor is 0.0019mm/bit (Table 3), and a stiffness of ±250N/mm the expected accuracy can be calculated: 250 · 0.0019 = 0.4750N. Which is within the required accuracy is 0.5N (Table 1).

Distance measurement by hall effect sensors (HONEYWELL S&H - SS495A - IC, HALL EFFECT SENSOR, LINEAR, TO-92-3) is performed by using a magnet (Neodymium disc magnet, Eclipes N802 diameter 3mm) that is placed at a distance from the hall effect sensor. By measuring the magnetic field density, the distance of the magnet relative to the sensor can be measured. Implementing this hall sensor in combination with the magnet at each outside corner of the bending beam, we can measure the distance changes due to the force and moment extended to the FMT. A test prototype consisting of two bending beams with a hall effect sensor was made and tested, there is a non-linear third degree polynomial function between the forces extended to the beam and the output voltage of the sensor. The hall sensor is placed in a prefabricated gap on each bending beam (Fig.13, Appendix C). The gap prevents direct contact between the magnet and the sensor when the system is loaded maximally. The hall effect sensors are connected to an extended LABJACK (LabJack U3-LV). The connection of all electronics and sensors can be found under Appendix E.

**Software**

A user interface was made to configure the settings of the sensors and gather the data from all sensors. A complete description of the interface is given in Appendix F. The data was gathered with MATLAB code. First, the setting of the LABJACK was configured by setting all LABJACK channels to be analogue inputs and initialisation of a streaming function to gain the voltage data. This streaming function stores the data with a specific sample frequency on a buffer inside the LABJACK. MATLAB gathers this voltage data, clears the buffer inside the LABJACK, calculates the corresponding forces, stores the data in a data matrix and plots the data inside the user interface. The main advantage of the streaming function is the ability to measure at high sample frequencies.

**Data Filtering**

All raw voltage data from the sensors were filtered by a low pass Butterworth filter at a cut-off frequency of 4Hz. This range is selected because it is unlikely that the human movement generates task features faster than 0.25 seconds [8]

### 3.4 Measurement protocol

#### 3.4.1 Calibration

The goal of this calibration step is to make a force voltage relationship for each sensor for each force or moment in each direction. With these data we can accurately calculate the moment and forces in all directions exerted on the force measurement table (FMT). When the forces and moments on the FMT can be determined, we can accurately calculate the position and direction of the actual force with the theoretical model consisting of Eq.(31 and 32).

The force measurement table (FMT) was placed on top of a steady robust table to prevent movement of the surroundings to influence the sensors. The FMT was fixated on the table. A spirit level was used to make sure the centre plate is exact in horizontal position.

Before calibration, all of bolts made for adjusting the distance between magnet and sensor, were set at the same output voltage at a distance of 3mm from the sensor. This was done by reading the input of the LABJACK into the computer and changing the magnet distance by hand.

In each direction, the force-voltage graph was determined between 0 and 100N in steps of 10N (Table 3). For each direction ten measurements were done. Also 0.5N difference at each load step was measured. The measurement data of each sensor were stored into the computer. Force was applied by weight pulley system. A stand with two pulleys connecting the weight with the centre of the centre plate with a wire. By adjusting the position of the stand, the direction of the force was changed. To apply a moment on the force measurement table, a beam was connected to the supporting system to make a moment arm. By applying a force, with the same pulley-stand system, at the end of the beam, a moment on the FMT was created.

During each measurement step, ten measurements during 1s of voltage data of each sensor was measured. The average voltage of those ten measurements was stored into the computer. A basic fitting tool in MATLAB was used to find the optimal 2nd order polynomial that fit the force-voltage data of each sensor in each direction. The 2nd order polynomial function fit was chosen because visual manually reviewed this was the most simple and correct fit for the data curve. Force-voltage curves of each sensor at their specific direc-

<table>
<thead>
<tr>
<th>Sensor</th>
<th>Accuracy</th>
<th>Costs</th>
<th>Maximal force</th>
<th>Time Delay</th>
</tr>
</thead>
<tbody>
<tr>
<td>Load Cell</td>
<td>0.25N/bit</td>
<td>280€</td>
<td>500N</td>
<td>Nil</td>
</tr>
<tr>
<td>Flexiforce</td>
<td>± 3%</td>
<td>280€</td>
<td>110N</td>
<td>5ms</td>
</tr>
<tr>
<td>Hall effect</td>
<td>0.0019mm/bit</td>
<td>130€</td>
<td>depended on stiffness</td>
<td>Nil</td>
</tr>
<tr>
<td>sensor</td>
<td>0.45 N/bit</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

### Table 2. An overview of the investigated sensors for force measurement.
two independent forces (systems, as used in the calibration step, were used to provide 3.4.2 Portal-tip forces
directions at the centre of the centre plate(Appendix G ).
ware code which calculate the exact force, moment and those
and stored in MATLAB, was used to create a MATLAB soft-
in time. The outcome of the polynomial systems, calculated
be sure that the force-voltage relationship does not change
bration measurements ware redone after a couple of days to
Table 3. The table shows at measurement steps done to calibrate
the force measurement table.

<table>
<thead>
<tr>
<th>Direction</th>
<th>Magnitude [N]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Force x</td>
<td>0 10 10.5 ... 100 100.5</td>
</tr>
<tr>
<td>Force y</td>
<td>0 10 10.5 ... 100 100.5</td>
</tr>
<tr>
<td>Force z</td>
<td>0 10 10.5 ... 100 100.5</td>
</tr>
<tr>
<td>Moment x</td>
<td>0 10 10.5 ... 100 100.5</td>
</tr>
<tr>
<td>Moment y</td>
<td>0 10 10.5 ... 100 100.5</td>
</tr>
<tr>
<td>Moment z</td>
<td>0 10 10.5 ... 100 100.5</td>
</tr>
</tbody>
</table>

tional force were stored as a polynomial system. The cali-
ration measurements ware redone after a couple of days to be sure that the force-voltage relationship does not change in time. The outcome of the polynomial systems, calculated and stored in MATLAB, was used to create a MATLAB soft-
ne code which calculate the exact force, moment and those directions at the centre of the centre plate(Appendix G ).
3.4.2 Portal-tip forces
The goal of this research was to prove the theoretical model described in Eq. (10 - 15). Two stands with a pulley systems, as used in the calibration step, were used to provide two independent forces \( F_{\text{port}}, F_{\text{tip}} \) in different directions. For multiple scenarios measurements were taken with different force distribution and directions between both forces (Table 4). First, a single force with direction was provided to the system. Secondly two forces with different directions were extended to the system.
A computer programming software code MATLAB was used to calculate the moments, forces and their directions at the centre of the centre plate at each measurement step using the polynomials determined in Table 2 (Appendix G). The direction of the portal force is chosen, as such that the force is perpendicular towards the distance vector calculated between \( r_{\text{port}} \) and \( r_{\text{tip}} \). A second MATLAB code was used to calculate the portal and tip forces with the use of the theoretical model (Appendix H). An example of calculating two forces form a the forces and moments measured on the table is given in Fig.6
Each scenario was measured during a time period of 5seconds and a sample frequency of 50Hz, form the calculated forces the standard deviation and mean value were calculated. All scenarios were compared with the outcome of the calculation from the MATLAB code.
3.4.3 Instrument-Arthroscope comparison.
The goal of this research step was to test the assumption, done in the theoretical model, that the instrument applies significant more force to the specimen than the arthroscope, during arthroscopic surgery. The PASSPORT v2, was used as phantom model of the human knee. Here for the Phantom part of the simulator is decoupled from its original frame and fixated to the supporting system. All sensors inside the passport knee will be neglected in this research. Only the data from the force measurement table was used. After connecting the PASSPORT phantom knee to the force measurement table, a new calibration step was done. All output voltage form the sensors were set to at zero force. In this way, all forces measured above the zero setting, could be explained by the forces extended to the PASSPORT system. Standard arthroscopic equipment with a 30°4mm arthroscope and a standard knee arthroscopic probe , was used for this measurement. Two intermediates were asked to perform a standard navigation round with the use of the arthroscope and a probing instrument on the PASSPORT phantom. Secondly, the intermediates were asked to perform the same navigation round, but only with the arthroscope inside the knee phantom. To prevent change in tactics, the intermediate was asked to hold the instrument in the right hand at the height of the knee phantom and do not use it to apply force to the knee. After that the same intermediate was asked to repeat both navigation rounds (with instrument and without instrument) ten times. The standard deviation and mean of the forces measured during both set of rounds are compared. If there is a significant lower standard deviation between only scope set and both instruments set, the assumption that the scope has significant less force extended to the specimen would be true.

Data Analysis Instrument-Arthroscope comparison
From the raw voltage data the force at each sensor is calculated with the polynomials of Table 5. At each measurement step the norm of the force vector is calculated (predicted) data values, is close to 1 at each regression. The correlation coefficient between the observed and modelled data at each navigation round the standard deviation and mean is calculated. The standard deviations of the only arthroscope rounds and the arthroscope with instrument rounds are compared with a students t-test.

4 Results
4.1 Calibration
In Table 5 the polynomials of the calibration are shown. The coefficient of determination, the square of the sample correlation coefficient between the observed and modelled (predicted) data values, is close to 1 at each regression. The difference between first and second calibration data was less than 1%.

4.2 Tip - Portal force
The results of the measurements on of tip-portal forces, are given in Appendix I. In Fig.9, a representation of the mean and standard deviation of the measurements on each scenario is given. The average error between predicted forces and actual measured forces is 0.32N for the portal force and 0.15N for the tip force. With an average standard deviation
of 0.11N of both portal and tip force measurements.

### 4.3 Instrument-Arthroscope comparison

In Fig.10A and Fig.10B two box-plots are shown, which show the results of the arthroscope-instrument comparison experiments. The box-plot of Fig.10A shows the mean value of both rounds, and Fig.10B shows the standard deviation in both rounds. The average mean value of the rounds with only the arthroscope is 2.4N with an average standard deviation of ±1.2N, for the rounds with arthroscope and instrument the average mean value and standard deviation is: 4.5N ± 2.4N. The t-test shows a significant difference between the forces for the task using only the arthroscope compared to the task using an instrument and the arthroscope for both mean value (p=4.2E-8) and standard deviation (p=3.19E-8).

### 5 Discussion

The most important finding is that the developed force measurement table is able to measure force accurately and within the set range performed in a simulated arthroscopic procedure. The required ±0.5N accuracy is reached: ±0.32N. The system was able to calculate to discriminate two forces and indicate their directions[Fig.10]. These force measurement table is able to resist and measure joint stress forces to 750N. The system was well tested in each direction with forces over 750N. This makes the system robust and well able to resist non cautious users.

The non-linear relationship between magnetic field and output voltage, make it possible to adjust the sensitivity of the hall effect sensors. Decreasing the magnet distance will increase sensitivity but decrease the overall range. This could be useful when the system is used for procedures with a lower range of force magnitudes, for instance during wrist
arthroscopic procedures.
The development of this sensor is clinically relevant because measuring the forces during arthroscopic cadaveric courses will add the possibility to provide objective force feedback to the trainee [1, 3, 4]. This helps to decrease learning time and patient safety.

Another user function of the FMT could be to measure forces during arthroscopic procedures. Since a few studies describing the forces during arthroscopic procedures were found [10, 11], there is a need this kind of research. For the implementation of providing force feedback during cadaveric training courses, data of the actual desired force ranges is needed. Also the resulting data acquired from these studies can be used for the assessment of other medical simulators. Calibration of the system could be done accurate by a 2nd order polynomial fit of the force voltage data. The calibration of the system is reliable after a couple of days. But there was a difference was noticed when the system has been reassembled. It is advisable to redo the calibration after reassembling the FMT. Although the FMT can accurate distinguish between two independent contact forces, the calculation relies also an accurate estimation of the position of the forces. In the experiments the position measurements were done with applied forces that remained static. When the position of the forces are dynamic, an accurate tracking system needs to be added to the FMT. For the tracking of the instruments several options are available tracking the rotational orientation of the instrument, the depth of the instrument, and the position of the portal in 3D space.
The theoretical model (Eqs.3-11 and 19-32) is based on a simplified mathematical model of the real life situation. This because we assumed the existence of only two contact forces...
Fig. 8. Two examples of the force data gained during a navigation round with scope and instrument (top) and with only the scope (bottom). The standard deviation and mean value during the specific round are shown as a dotted line.

Table 5. Table showing the polynomials calculating the forces corresponding to the voltage measured at each sensor and the coefficient of determination.

<table>
<thead>
<tr>
<th>Sens</th>
<th>Dir</th>
<th>Polynomial</th>
<th>( R^2 )</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>y</td>
<td>( p_1 = -752, p_2 = 2706, p_3 = -2401 )</td>
<td>0.9993</td>
</tr>
<tr>
<td>2</td>
<td>y</td>
<td>( p_1 = -965, p_2 = 3391, p_3 = -2941 )</td>
<td>0.9986</td>
</tr>
<tr>
<td>3</td>
<td>y</td>
<td>( p_1 = -6462, p_2 = -20350, p_3 = 15960 )</td>
<td>0.9996</td>
</tr>
<tr>
<td>4</td>
<td>y</td>
<td>( p_1 = 2555, p_2 = -8500, p_3 = 6997 )</td>
<td>0.9997</td>
</tr>
<tr>
<td>5</td>
<td>x</td>
<td>( p_1 = -876, p_2 = 3156, p_3 = -2822 )</td>
<td>0.9995</td>
</tr>
<tr>
<td>6</td>
<td>x</td>
<td>( p_1 = 1656, p_2 = -5739, p_3 = -4891 )</td>
<td>0.9973</td>
</tr>
<tr>
<td>7</td>
<td>x</td>
<td>( p_1 = 790.9, p_2 = -2845, p_3 = -2507 )</td>
<td>0.9993</td>
</tr>
<tr>
<td>8</td>
<td>x</td>
<td>( p_1 = -2022, p_2 = 6006, p_3 = -5380 )</td>
<td>0.9996</td>
</tr>
<tr>
<td>9</td>
<td>z</td>
<td>( p_1 = -364, p_2 = 1438, p_3 = -1363 )</td>
<td>0.9999</td>
</tr>
<tr>
<td>10</td>
<td>z</td>
<td>( p_1 = -434, p_2 = 1670, p_3 = -1552 )</td>
<td>0.9996</td>
</tr>
<tr>
<td>11</td>
<td>z</td>
<td>( p_1 = -422, p_2 = 1634, p_3 = -1552 )</td>
<td>0.9997</td>
</tr>
<tr>
<td>12</td>
<td>z</td>
<td>( p_1 = -1512, p_2 = 5164, p_3 = -4353 )</td>
<td>0.9838</td>
</tr>
</tbody>
</table>

Fig. 10. Box-plots of the Mean value (10A) and the standard deviation (10B) of all rounds with only arthroscope (scope) and with both arthroscope and instrument (instr). Each box-plot consist of 20 measurements, 10 of each subject.

6 Conclusions

A 6DOF force platform, the Force Measurement Table (FMT), to measure force and torque in three direction during training of arthroscopic surgery on cadaveric specimen was developed. The FMT will make it possible to implement objective force feedback to the trainee during cadaveric training courses without adjustment of the instrumentation. A low cost design with cheap sensors were used to measure a broad range of forces 0 - 750N with an acceptable accuracy of ±0.32N. A theoretical model can distinguish between two independent contact forces form the measured forces and the positions of the contact forces. A pilot study towards the difference in forces exerted by the instrument or the arthroscope, shows a significant less force of only the arthroscope relative to the use of both instruments.

Acknowledgements

The Author would like to thank the two intermediates who were participating during the experiments. Furthermore, the Technicians of the Delft University of Technology 3ME workshop, for helping and advising building the FMT. Special thanks to this Master Thesis Supervisors: Gabrielle Tu-
jthof and Tim Horeman.

References
[13] Clausen, C. E., McConville, J. T., and Young, J. W., 1969. Weight, volume, and center of mass of segments of the human body. Aerospace Medical Research Labo-
Appendix A: Forces during arthroscopy

An overview of the in literature found dimensions and forces during arthroscopic procedures are shown in Table 7 and Table 6.

Table 6. Forces during arthroscopy.

<table>
<thead>
<tr>
<th>Joint</th>
<th>Gravitational forces</th>
<th>Instrument forces</th>
<th>Joint stress forces</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle</td>
<td>0.70-1.2 kg [13]</td>
<td>unknown</td>
<td>80-150 N [14]</td>
</tr>
<tr>
<td>Knee</td>
<td>6-12 kg [13]</td>
<td>1.3-5.5 N [10]</td>
<td>87-142 N [16]</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>12.7±6.8 MPa [13]</td>
</tr>
<tr>
<td>Wrist</td>
<td>0.3-0.7 kg [13]</td>
<td>unknown</td>
<td>40-80 N [15]</td>
</tr>
<tr>
<td>Hip</td>
<td>6-12 kg [13]</td>
<td>unknown</td>
<td>100-300 N [17]</td>
</tr>
</tbody>
</table>

Table 7. Fixation of the specimen:

<table>
<thead>
<tr>
<th>Joint</th>
<th>Joint stress force</th>
<th>Fixed bone</th>
<th>Shape of bone</th>
<th>Size</th>
<th>Stress applied bone</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle</td>
<td>100-300 N</td>
<td>Tibia</td>
<td>Cylindrical</td>
<td>Diameter: 20-30 mm, Length: 350-380 mm</td>
<td>foot</td>
</tr>
<tr>
<td>Elbow</td>
<td>0 N</td>
<td>Humerus</td>
<td>Cylindrical</td>
<td>Diameter: 12-20 mm, Length: 300-350 mm</td>
<td>none</td>
</tr>
<tr>
<td>Knee</td>
<td>87-142 N</td>
<td>Femur</td>
<td>Cylindrical</td>
<td>Diameter: 20-30 mm, Length: 400-450 mm</td>
<td>tibia</td>
</tr>
<tr>
<td>Wrist</td>
<td>40-80 N</td>
<td>Radius Ulna</td>
<td>Cylindrical</td>
<td>Diameter: 10-20 mm, Length: 230-280 mm</td>
<td>hand</td>
</tr>
<tr>
<td>Hip</td>
<td>100-300 N</td>
<td>Ilnimate</td>
<td>Flat round</td>
<td>Thickness: 28 mm</td>
<td>femur</td>
</tr>
<tr>
<td>Shoulder</td>
<td>50 N</td>
<td>Scapula</td>
<td>Flat round</td>
<td>Thickness: 28 mm</td>
<td>Humerus</td>
</tr>
</tbody>
</table>
Appendix B: Report instrument movement analysis

Introduction

In order to test the assumption that the arthroscope movement is less than the instrument movement during an arthroscopic knee surgery, a data analysis on the 2D movement data from the PASSPORT arthroscopic knee simulator performed. In previous experiments data about the two dimensional position of both the arthroscope and probe were gathered. During this experiment, the novices, intermediates and experts were asked to perform a specific navigation task. During this task, the two dimensional position of the probe and arthroscope were tracked with a webcam and software recognized one marker per instrument, assuming that the portal position was constant in time. These processed 2D-markers position were saved in a data log. Simultaneously, the forces on the femur and tibia bone, the flexion angle of the leg and the varus and valgus joint stress distance were stored in this data log. For the data analysis, the position data of the cope and probe and the force data of both bones were used to analyse. As it was expected that experts outperformed the novices and intermediates, only the data from experts was used to answer our main question.

Method

Data

The data was gathered from a constructed .mat-file where per surgeon four data matrixes were included. The function: 
\[ [t, \text{xn1,yn1,xm2,ym2, Ffem,Ftib}, \text{data}] = \text{getdata(dataname)} \], returns the data matrix (data), the time vector (t), position of two marker arrays (xm1, ym1, xm2, ym2) and the force vectors (Ffem, Ftib). The input is a string with the name of the data log. Simultaneously, the forces on the femur and tibia bone, the flexion angle of the leg and the varus and valgus joint stress distance were stored in this data log. For the data analysis, the position data of the cope and probe and the force data of both bones were used to analyse. As it was expected that experts outperformed the novices and intermediates, only the data from experts was used to answer our main question.

Data filtering

The position data had imperfections, that were corrected with the MATLAB function: 
\[ [\text{fdata}, \text{IArray}] = \text{filterdata(data, maxv, maxdx)} \]. With inputs the data matrix gathered form the data log (data). This filterdata m-file processed the following steps to correct the imperfections:

1. First of all when a marker was not detected in the webcam screen, a value of 4900 was written to the data. This is not a true tracking value, so with the analysis code it was corrected to remove every value above 2000 of the position data (MATLAB function: \[ \text{data} = \text{removelostdata(data)} \]).

2. Second, the marker sequence was not indicated to be arthroscope or probe, but the marker number indicates which marker was first in the screen. The function \[ \text{[xm1,ym1,xm2,ym2] = leftrightcheck(xm1,ym1,xm2,ym2)} \], looks at the mean value of the xm1 and xm2 position. As the task was performed on a right knee with the Arthroscope in the lateral portal, we know that the highest mean value of the marker array should indicated the arthroscope and therefore should be right in the webcam screen and indicated to be the Arthroscope. If the mean value of xm1 is not the highest mean value the arrays will be changed. After the use of this function the arrays of marker 1 (m1) is supposed to be the arthroscope and the array of marker 2 (m2) is supposed to be the probe.

3. Third, the markers positions could be very close to each other, thus would make step 2 difficult to carried. Function: 
\[ \text{[xm1,ym1,xm2,ym2,IArray,tel] = samemarkercheck(xm1,ym1,xm2,ym2,IArray,tel)} \], determines if the markers are in almost the same position. If this is the case, the x-values of both markers, then the x- and y-value are assigned to the marker with the smallest difference between the previous known x-position and the current x-position. The other value will be exchanged with previous known position values.

4. Fourth, when a marker has a high position derivative (>20 pixels per sample time), it can indicate that a the tracking software has tracked another yellow zone than the marker. The function: \[ \text{[xm1,ym1,xm2,ym2] = lostmarker(xm1,ym1,xm2,ym2)} \], determines if a marker has high derivative value, a sudden change of position of the marker position is closer to the last known position of the other marker than the x-value of the position of the first marker. If this is true, the x- and y-value of the markers will be exchanged. And the first marker will be removed.

5. Fifth, if a marker does change suddenly, high value of the position derivative (>20 pixels per sample time), this could indicate that a the tracking software has tracked another yellow zone than the marker. The function: 
\[ \text{[xm1,ym1,xm2,ym2] = removehighdisp(xm1,ym1,xm2,ym2, maxy)} \], determines if a marker has high derivative value, a sudden change of position. If so the marker position value will be removed from the array and changed in to a NaN (Not A Number).

6. After all filtering steps the data still changes the position of both markers, therefore an additional step is taken to filter marker exchanges. The function: 
\[ \text{[xm1,ym1,xm2,ym2] = checktomean(xm1,ym1,xm2,ym2,maxy)} \], determines to which mean value of the other marker, the x- and y-value of both markers will be changed.
Fig. 11. Graphs of the data for one expert at test sample 4. Left graphs unfiltered data, right graphs filtered data. Upper graphs x-position of the instrument and arthroscope, middle graph velocity of the instrument and arthroscope and the lower graphs the forces on the tibia and femur bone. The vertical lines shows the border between the different parts: lateral, midsection and medial side of the knee.

Plotting the data

The file read_2D.m and read_2D_delen.m gets the data from the data file, filters the data and plot them into a graph. The read_2D_delen.m file does read the data which is split up into three parts, corresponding with the lateral side, midsection and the medial side of the knee, respectively. Part one consist of indicating landmarks:1,2,3,4, part two consist landmark 5 and part three consist of landmarks: 6,7,8 and 9 (Fig.1) The read_2D_delen.m file plots the graph, with on the left side the unfiltered data, on the right side the filtered data. The upper graph shows the position of the x-value of both arthroscope and probe, the middle graph shows the derivative of the upper graph, representing the instrument velocity and the lower graph shows the total combined forces measured on the tibia and femur bone(Fig.11).

Data analysis

For each expert the standard deviation of the position vector ($pos_{vec} = \sqrt{x^2 + y^2}$) at each of the three parts for the Arthroscope and the probe. Combining those results we analysed the standard deviation with a students t-test ($P<0.05$). In Fig.12, Fig.13 and Fig.14 the results of the standard deviation is shown including the p-value results of the t-test at each part of the navigation round.

Results

At part one and three, the lateral and medial side of the inspection round, we can conclude that the Arthroscope significantly moves less than the probe. In part two, the anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL) inspection, the arthroscope significantly deviate more than the probe.

Discussion

During the first part and the last part, the arthroscope almost keeps the same position. The arthroscope will only be moved slightly to have the another view of the compartment. The movement is mostly done by the probe, inspecting the condyle, the tibia plateau and the meniscus. This is shown in Fig.12 and Fig.14, where the arthroscope has significantly less movement than the probe. In the first part, the lateral side, there is a high variance in the deviation of the probe. This could be explained by the fact that the probe is inserted into the joint during this measurement. More movement is needed to orientate the probe in the desired position. In the midsection, the arthroscope has more deviation than the probe, this could be explained by the fact that during the inspection of the ACL and PCL, the surgeon moves the arthroscope form the lateral
side to the medial side. Also both Arthroscope and probe has more movement deviation during this part compared with the other parts, this is also due the fact that both instruments are moved from lateral to medial side.

Fig. 12. Box-plot of the standard deviation in part one of the navigation round. Four experts with each four measurements were included. ($p = 0.0072$)

Fig. 13. Box-plot of the standard deviation in part two of the navigation round. Four experts with each four measurements were included. ($p = 0.2924 \times 10^{-5}$)

Fig. 14. Box-plot of the standard deviation in part three of the navigation round. Four experts with each four measurements were included. ($p = 0.0148 \times 10^{-3}$)
Appendix C: Bending beams

The bending beam (Fig. 15) can be modelled as a beam with on one side an all DOF fixation and the other side a sliding fixation, allowing free translation along the vertical plane. Then the displacement and angle due to a force can be described as:

\[
\theta = -\frac{FL^2}{2EI} + \frac{ML}{EI} = 0 \tag{33}
\]

\[
\delta = -\frac{FL^3}{3EI} + \frac{ML^2}{2EI} \tag{34}
\]

Since the beam is fixated with a sliding fixation, the angle rotation of the force will me zero (Eqs. 33, 34). Then the moment on the sliding fixation can be calculated as:

\[
M = \frac{FL}{2} \tag{35}
\]

And so is the displacement on due to a force:

\[
\delta = -\frac{FL^3}{3EI} + \frac{FL^3}{4EI} \tag{36}
\]

\[
= -\frac{1}{12} \cdot \frac{FL^3}{EI}
\]

The inertia can be calculated in both x and y-direction, where in the x-direction the bending beam is most stiff and in the y-direction the bending beam is less stiff.

\[
I_x = -\frac{1}{12} \cdot (y - d) \cdot x^3 \tag{37}
\]

\[
I_y = -\frac{1}{12} \cdot x \cdot (y^3 - d^3) \tag{38}
\]

In this equation \(x\) is the length of the beam in the x-direction, \(y\) is the length of the beam in y-direction and \(d\) is the height of the cut. The displacements in both x and y-direction can be calculated with Eqs. (36-38):
\[ \delta_x = \frac{F L^3}{E \cdot (y - d) \cdot x^3} \]  
\[ \delta_y = \frac{F L^3}{E \cdot (y^3 - d^3) \cdot x} \]  

Then we can calculate the stiffness in each direction:

\[ k = \frac{F}{\delta} \]  
\[ k_x = \frac{E \cdot (y - d) \cdot x^3}{L^3} \]  
\[ k_y = \frac{E \cdot (y^3 - d^3) \cdot x}{L^3} \]

In Eqs. 1 and 2 the maximum force allowed on the system was calculated. With \( r = 0.16m \), the maximum force in one direction is 933N. Divided by the number of four beams in one direction that is the maximum force load on each beam \( \frac{933}{4} = 233N \):

The maximum allowed displacement on a beam: \( \delta_{\text{max}} = 3mm \). This is depended on the maximum distance between hall sensor and magnet (Appendix). Thus we are looking for a beam with stiffness:

\[ k_y = \frac{F}{\delta} = \frac{233}{3} = 78N/mm; \]  

The chosen dimensions of the beam are shown in Table 8.

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Description</th>
<th>Dimension</th>
</tr>
</thead>
<tbody>
<tr>
<td>( L )</td>
<td>Length of the cut through the beam</td>
<td>50mm</td>
</tr>
<tr>
<td>( d )</td>
<td>Height of the cut</td>
<td>7mm</td>
</tr>
<tr>
<td>( y )</td>
<td>Height of the beam</td>
<td>10mm</td>
</tr>
<tr>
<td>( x )</td>
<td>Width of the beam</td>
<td>15mm</td>
</tr>
<tr>
<td>( E )</td>
<td>Modulus of Elasticity</td>
<td>71.7 Gpa</td>
</tr>
</tbody>
</table>

Than \( k_x \) and \( k_y \) can be calculated with Eqs. 42 and 43:

\[ k_x = \frac{E \cdot (y - d) \cdot x^3}{L^3} = 2.12E^3N/mm \]  
\[ k_y = \frac{E \cdot (y^3 - d^3) \cdot x}{L^3} = 85N/mm \]

Which is a little bit more stiff as required, but this makes the table more robust. The actual stiffness depends on more calculations than this simplified model.
Appendix D: Test prototype

To test the hall sensor and the bending effect of the bending beams a test prototype consisting of two bending beams was placed as in the xy-plane of the full prototype was made. Hereby we could setup a relationship between force and distance and the output voltage of the sensor. The relation between distance and output voltage is a non-linear parabolic function (Fig. 16). At small distance (<3mm) the sensor is more sensitive to changes in distance than at large distances. Because the relation between the distance between sensor an magnet and the force extended to the table is linear elastic, the relationship between force and output voltage is also a non-linear third degree polynomial function (Fig. 17). The maximal expected range of the hall-effect sensor in distance is between 1 and 5mm. A 3th order polynomial fit with MATLAB basic fitting toolbox describes the relation between voltage and sensor distance as:

\[ U_{\text{sensor}} = 5.8E^{-6} \cdot d^3 - 0.13E^{-3} \cdot d^2 + 9.7E^{-3} \cdot d + 2.8 \]  

(48)
Appendix E: Sensors and Electrical connections

In fig. 19 an overview is given of the Electrical connections. The Sensors are connected to an electrical circuit (fig. 20) to give them a steady 5V power source and translate the 0-5V output range to a 0-2.4V range. This because the LABJACK requires this range.

Fig. 18. Picture with a sketch of the electronical connections of the FMT.

Fig. 19. A schematic drawing of the electronically circuit. On top the 2 of the 12 connectors are shown. A 7V power source is concerted to a 5V source. The connected output is concerted form a 0-5V range to a 0.2.4V range by a variable resistance.
Appendix F: Software interface

An interface was made and designed with the GUIDE toolbox in MATLAB (Fig. 20). The interface is designed to make measurements of all sensors and calculate the corresponding forces during a predefined time period. With the sensor 1 to 12 radio buttons, the specific sensors can be turned on or off. The listboxes next to the radiobutton determine on which port of the Labjack the sensor is connected. At the tend textbox the length of the measurement can be adjusted. The data can be saved to a specific file named as typed into the filename textbox, when the save data? radiobutton is turned on. The measurement can be started and stopped with the start and stop button. When the user want to see the data during the measurement, he need to turn on the show data during measurement radiobutton. Before the measurement is started, the a zero measurement can be done by pressing the set zero forces button. This will all forces at the measured on the table to be zero. When the measurement is started, the voltage on each sensor together with their corresponding forces is shown in the voltage and forces columns. Also in the first graph the voltage is on each sensor at each time step is shown, in the second graph the corresponding forces at each sensor at each time step is shown and in the third graph the calculated directional forces (Fx, Fy, Fz) form the sensor forces is shown. The corresponding time since the start of the measurement is shown in the time textbox.

Fig. 20. User interface used to control measurements.
Appendix G: Force vector calculation code

MATLAB code calculating the forces and moments at the measurement table from the voltage data and the force voltage polynomials:

```matlab
function [F_vec] = calc_forcevector(V0,V)
% vector size n = the number of samples in the voltage vector
% and V is the total of sensors activated.
[n,m] = size(V);

% get the calibrated polynomials
load('poly_sensoren.mat')

% calculate the forces corresponding to the predefined voltages when the % all forces are zero.
F0(1) = s1_poly(V0(1));
F0(2) = s2_poly(V0(2));
F0(3) = s3_poly(V0(3));
F0(4) = s4_poly(V0(4));
F0(5) = s5_poly(V0(5));
F0(6) = s6_poly(V0(6));
F0(7) = s7_poly(V0(7));
F0(8) = s8_poly(V0(8));
F0(9) = s9_poly(V0(9));
F0(10) = s10_poly(V0(10));
F0(11) = s11_poly(V0(11));
F0(12) = s12_poly(V0(12));

% for all samples
for i=1:n
% calculate the force corresponding to the voltage at each sensor
F_sensoren(i,1) = s1_poly(V(i,1))-F0(1);
F_sensoren(i,2) = s2_poly(V(i,2))-F0(2);
F_sensoren(i,3) = s3_poly(V(i,3))-F0(3);
F_sensoren(i,4) = s4_poly(V(i,4))-F0(4);
F_sensoren(i,5) = s5_poly(V(i,5))-F0(5);
F_sensoren(i,6) = s6_poly(V(i,6))-F0(6);
F_sensoren(i,7) = s7_poly(V(i,7))-F0(7);
F_sensoren(i,8) = s8_poly(V(i,8))-F0(8);
F_sensoren(i,9) = s9_poly(V(i,9))-F0(9);
F_sensoren(i,10) = s10_poly(V(i,10))-F0(10);
F_sensoren(i,11) = s11_poly(V(i,11))-F0(11);
F_sensoren(i,12) = s12_poly(V(i,12))-F0(12);

% calculate overall force on the FMT
Fy(i)= 2*(F_sensoren(i,1)+F_sensoren(i,2))+2.5;
Fx(i)= F_sensoren(i,5)+F_sensoren(i,6)+F_sensoren(i,7)+F_sensoren(i,8)-0.5;
Fz(i)= F_sensoren(i,9)+F_sensoren(i,10)+F_sensoren(i,11)+F_sensoren(i,12)+0.8;

% calculate the moments on the FMT
Mx(i) = (F_sensoren(i,9) + F_sensoren(i,10) - F_sensoren(i,11) - F_sensoren(i,12))*130;
My(i) = (F_sensoren(i,9) - F_sensoren(i,10) - F_sensoren(i,11) + F_sensoren(i,12))*150 - 32*Fz(i);
Mz(i) = (-F_sensoren(i,5) - F_sensoren(i,6) + F_sensoren(i,7) + F_sensoren(i,8))*100*2;

end
% make a complete force vector
F_vec= [Fx' Fy' Fz' Mx' My' Mz'];
```
Appendix H: Portal and Tip Force calculation code

MATLAB code calculating the tip force and portal forces form the moments and forces on the FMT and the position of the tip and the portal using the theoretical model (Eqs.(31),(32):

```matlab
function [Fport, Ftip] = calc_instrF(F_vec,r_port,r_tip)
% n is the number of samples to be calculated
% m is 6, Fx, Fy, Fz, Mx, My, Mz
[n,m] = size(F_vec);

%for al samples:
for i=1:n
    %indicate force and moment vector
    Fs = F_vec(i,1:3);
    Ms = F_vec(i,4:6);
    %calculate virtual force and moment on handle
    vFh = Fs;
    vMh = -cross(Fs,r_port) - Ms;
    % calculate distance vector between portal and tip
    vb= r_tip-r_port;
    % calculate exact distance between portal and tip
    b= norm(vb);
    % calculate angular rotations between the global coordinates and
    %the instrument coordinates
    if vb(3)==0 && abs(vb(2))>0
        phi=0.5*pi;
    else
        phi = +atan(vb(2)/vb(3));
    end
    if vb(3)==0 && abs(vb(1))>0
        tet=-0.5*pi;
    else
        tet = -atan(vb(1)/vb(3));
    end
    if vb(2) == 0 && abs(vb(1))>0
        psi=-0.5*pi;
    else
        psi = -atan(vb(1)/vb(2));
    end
    %calculate the rotational matrix
    R = rotmatrix(phi,tet,psi);
    %calculate the forces and moments on the virtual handle in
    %instrument coordinates
    vMhi = vMh*R;
    vFhi = vFh*R;
    %calculate portal and tip forces
    Fti=[(vMhi(2))/b;
         (vMhi(1))/b;
         vFhi(3)];
    Fpi =[(vMhi(2) + vFhi(1)*b)/b;
          (vMhi(1) + vFhi(2)*b )/b;
          0];
    %make a tip an portal force vector for each sample
    Ftip(i,:)=Fti';
    Fport(i,:)=Fpi';
end
```
Appendix I: Data table of the tip and portal force experiment

Table 9. Table showing the results of the tip-portal force measurements. The calculated tip and portal forces are given and the error between the actual tip or portal force and the calculated force is given.

<table>
<thead>
<tr>
<th>Scenario</th>
<th>$F_{port}$</th>
<th>$F_{tip}$</th>
<th>Calc $F_{port}$ Mean [N]</th>
<th>SD [N]</th>
<th>Calc $F_{tip}$ Mean [N]</th>
<th>SD [N]</th>
<th>$F_{port}$ - calc $F_{port}$ [N]</th>
<th>$F_{tip}$ - calc $F_{tip}$ [N]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.</td>
<td>5N</td>
<td>0N</td>
<td>4.65</td>
<td>0.10</td>
<td>-0.24</td>
<td>0.10</td>
<td>0.35</td>
<td>-0.24</td>
</tr>
<tr>
<td>2.</td>
<td>5N</td>
<td>0N</td>
<td>4.71</td>
<td>0.12</td>
<td>-0.02</td>
<td>0.05</td>
<td>0.29</td>
<td>0.02</td>
</tr>
<tr>
<td>3.</td>
<td>5N</td>
<td>0N</td>
<td>4.69</td>
<td>0.09</td>
<td>0.23</td>
<td>0.14</td>
<td>0.31</td>
<td>-0.23</td>
</tr>
<tr>
<td>4.</td>
<td>5N</td>
<td>0N</td>
<td>4.89</td>
<td>0.11</td>
<td>0.11</td>
<td>0.13</td>
<td>0.11</td>
<td>-0.11</td>
</tr>
<tr>
<td>5.</td>
<td>5N</td>
<td>0N</td>
<td>4.79</td>
<td>0.11</td>
<td>0.29</td>
<td>0.13</td>
<td>0.21</td>
<td>-0.29</td>
</tr>
<tr>
<td>6.</td>
<td>5N</td>
<td>0N</td>
<td>4.76</td>
<td>0.14</td>
<td>0.19</td>
<td>0.06</td>
<td>0.24</td>
<td>-0.19</td>
</tr>
<tr>
<td>7.</td>
<td>5N</td>
<td>3N</td>
<td>4.77</td>
<td>0.10</td>
<td>2.63</td>
<td>0.15</td>
<td>0.23</td>
<td>0.37</td>
</tr>
<tr>
<td>8.</td>
<td>5N</td>
<td>3N</td>
<td>4.60</td>
<td>0.15</td>
<td>2.85</td>
<td>0.13</td>
<td>0.40</td>
<td>0.15</td>
</tr>
<tr>
<td>9.</td>
<td>5N</td>
<td>3N</td>
<td>4.77</td>
<td>0.08</td>
<td>2.87</td>
<td>0.24</td>
<td>0.23</td>
<td>0.13</td>
</tr>
<tr>
<td>10.</td>
<td>5N</td>
<td>3N</td>
<td>4.67</td>
<td>0.08</td>
<td>2.81</td>
<td>0.15</td>
<td>0.33</td>
<td>0.19</td>
</tr>
<tr>
<td>11.</td>
<td>5N</td>
<td>3N</td>
<td>4.50</td>
<td>0.17</td>
<td>3.03</td>
<td>0.16</td>
<td>0.50</td>
<td>-0.03</td>
</tr>
<tr>
<td>12.</td>
<td>5N</td>
<td>3N</td>
<td>4.48</td>
<td>0.05</td>
<td>2.89</td>
<td>0.09</td>
<td>0.52</td>
<td>0.11</td>
</tr>
<tr>
<td>13.</td>
<td>5N</td>
<td>3N</td>
<td>4.77</td>
<td>0.08</td>
<td>2.94</td>
<td>0.01</td>
<td>0.23</td>
<td>0.06</td>
</tr>
<tr>
<td>14.</td>
<td>5N</td>
<td>3N</td>
<td>4.64</td>
<td>0.13</td>
<td>2.84</td>
<td>0.07</td>
<td>0.37</td>
<td>0.16</td>
</tr>
<tr>
<td>15.</td>
<td>5N</td>
<td>3N</td>
<td>4.70</td>
<td>0.09</td>
<td>2.92</td>
<td>0.04</td>
<td>0.30</td>
<td>0.08</td>
</tr>
<tr>
<td>16.</td>
<td>5N</td>
<td>3N</td>
<td>4.67</td>
<td>0.15</td>
<td>2.65</td>
<td>0.06</td>
<td>0.33</td>
<td>0.35</td>
</tr>
<tr>
<td>17.</td>
<td>5N</td>
<td>3N</td>
<td>4.60</td>
<td>0.13</td>
<td>3.01</td>
<td>0.12</td>
<td>0.40</td>
<td>-0.01</td>
</tr>
<tr>
<td>18.</td>
<td>5N</td>
<td>3N</td>
<td>4.65</td>
<td>0.09</td>
<td>2.96</td>
<td>0.06</td>
<td>0.35</td>
<td>0.04</td>
</tr>
<tr>
<td>Average:</td>
<td>-</td>
<td>-</td>
<td>4.68</td>
<td>0.11</td>
<td>0.17 - 2.87</td>
<td>0.11</td>
<td>0.32</td>
<td>0.15</td>
</tr>
</tbody>
</table>
A Appendix J: Rotational matrix

Here the Rotational matrix will be reduced. Rotational matrix of a single rotation about an axis of the coordinate system can be determined. From Fig. 21 we can describe:

\[ r_y' = r_y \cos \phi - r_z \sin \phi \]  \hspace{1cm} (49)

\[ r_z' = r_y \sin \phi - r_z \cos \phi \]  \hspace{1cm} (50)

where \( r_y' \) and \( r_z' \) are the projections of the vector \( r \) on the \( y_s, z_s \) axes (global coordinate system); \( r_y \) and \( r_z \) are the projections on the \( y_i, z_i \) axes (Local coordinate system on the instrument); \( \phi \) is the angle of rotation around the shared \( x \)-axis of the coordinate systems.

\[ r_s = r_y \cos \phi - r_z \sin \phi \]  \hspace{1cm} (51)

\[ r_s = r_y \sin \phi - r_z \cos \phi \]  \hspace{1cm} (52)

When the global coordinate system is defined by \( x_s, y_s, z_s \) and the local coordinate system \( x_i, y_i, z_i \) it follows that the rotation matrix around the \( x \)-axis is:

\[ R_x = \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos \phi & -\sin \phi \\ 0 & \sin \phi & \cos \phi \end{pmatrix} \]  \hspace{1cm} (52)

And for the rotation angles of \( \theta \) and \( \psi \) around respective the \( Y \) and \( Z \) axis, give the Rotational matrices:

\[ R_y = \begin{pmatrix} \cos \theta & 0 & \sin \theta \\ 0 & 1 & 0 \\ -\sin \theta & 0 & \cos \theta \end{pmatrix} \]  \hspace{1cm} (53)

\[ R_z = \begin{pmatrix} \cos \psi & -\sin \psi & 0 \\ \sin \psi & \cos \psi & 0 \\ 0 & 0 & 1 \end{pmatrix} \]  \hspace{1cm} (53)

Fig. 21. The projection of the vector \( r \) changes when the coordinate system is rotated around the \( X \) axis with an angle \( \phi \).

When the global coordinate system is defined by \( x_s, y_s, z_s \) and the local coordinate system \( x_i, y_i, z_i \) it follows that the rotation matrix around the \( x \)-axis is:

\[ R_x = \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos \phi & -\sin \phi \\ 0 & \sin \phi & \cos \phi \end{pmatrix} \]  \hspace{1cm} (52)

And for the rotation angles of \( \theta \) and \( \psi \) around respective the \( Y \) and \( Z \) axis, give the Rotational matrices:

\[ R_y = \begin{pmatrix} \cos \theta & 0 & \sin \theta \\ 0 & 1 & 0 \\ -\sin \theta & 0 & \cos \theta \end{pmatrix} \]  \hspace{1cm} (53)

\[ R_z = \begin{pmatrix} \cos \psi & -\sin \psi & 0 \\ \sin \psi & \cos \psi & 0 \\ 0 & 0 & 1 \end{pmatrix} \]  \hspace{1cm} (53)
The Rotational matrix, converting local to global coordinates, can be described as:

\[
R^{is} = R_x \cdot R_y \cdot R_z = \\
\begin{pmatrix}
\cos \phi \cdot \cos \theta & -\cos \theta \cdot \sin \psi & \sin \theta \\
\cos \theta \cdot \sin \psi + \sin \phi \cdot \cos \psi \cdot \sin \theta & \cos \phi \cdot \sin \psi - \sin \phi \cdot \sin \theta & -\cos \theta \cdot \sin \phi \\
\sin \psi \cdot \sin \phi - \cos \phi \cdot \cos \psi \cdot \sin \theta & \cos \psi \cdot \sin \phi + \cos \phi \cdot \sin \psi \cdot \sin \theta & \cos \phi \cdot \cos \theta
\end{pmatrix}
\] (54)
B Appendix K: Technical drawings of the FMT
4x M4x0.7 mag verzonken worden
2x 4x Ø 4.3 beide kanten verzonken

4x 2x Ø 4.3 verzonken m4

WEIGHT:

A3

SHEET 1 OF 1

SCALE: 1:5

DIMENSIONS ARE IN MILLIMETERS
SURFACE FINISH:
TOLERANCES:
   LINEAR:
   ANGULAR:

Q.A
MFG
APPV'D
CHK'D
DRAWN
R2,5max R5
15
5
20
200
215
2x 4x M4x0.7
beide kanten
voor verzonken imbuskop 0.8
22.5
22.5
215
220
200
20
5
4x M4x0.7
debie kantten
2x 4x M4x0.7
debie kanten
**DIMENSIONS ARE IN MILLIMETERS**

**SURFACE FINISH:**

**TOLERANCES:**

**LINEAR:**

**ANGULAR:**

**Q.A**

**MFG**

**APPV'D**

**CHK'D**

**DRAWN**
2x M4x0.7 niet verzonken

2x Ø 4.3 verzonken

36