An improvement for the scanning process in high accuracy head tracking

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Abstract:

Optical tracking systems have become state-of-the-art in radiation therapy. Nevertheless, they still cannot cope with the properties of soft tissue. We are developing a non-invasive method to automatically detect the cranial bone's surface using only infrared laser light. Here, we present simulations to describe the influence of increased incident angles of the laser ray on the forehead's surface caused by replacing the original robotic setup by galvanometric scanning mirrors. We show that the usable area on the forehead (angles below 45°) will decrease, but that successful scanning is still possible when the scanning distance is increased from 300 to 500 mm. These results are confirmed when comparing optical scans acquired with the robotic setup and the scanning mirror assembly. Additionally, we could determine the needed depth of field for three subjects (based on high-resolution MRI scans). At a distance of 500 mm, it ranges from 37 to 86 mm.

Keywords: tracking system, incident angles, depth of field, accuracy, radiation therapy, cranial

1 Introduction

Patient motion during treatment is the main source of error in cranial radiation therapy. Basically, there are two ways to handle this problem: First, the patient is immobilized by stereotactic frames or thermoplastic masks [1]. Second, tracking systems like 6D skull (Accuray, Inc.), OBI (Varian Medical Systems, Inc.) or Calypso (Varian Medical Systems, Inc.) monitor the position of the patient [2]–[4]. Thereby, stereotactic frames are the most accurate but also the most expensive and most uncomfortable solution. Thermoplastic masks are a tradeoff between accuracy and comfort for the patient. Tracking systems using X-rays suffer from low sampling rates and further radiation dose for the patient. On the other hand, state-of-the-art optical tracking systems are inaccurate due to the properties of soft tissue (reflection, absorption, scattering, deformation) [5]–[7]. To overcome these problems, we are currently developing an optical tracking device for cranial radiation therapy which takes these properties into account. Initial Monte Carlo simulations of interaction between light soft tissue showed that it is possible to extract optical features [8] which correlate with the skin thickness. Using these features, the accuracy of optical tracking systems can be improved by a combination of a surface scans with depth information, enabling us to non-invasively track the cranial bone using infrared laser light only.
For this purpose we have built a robot-mounted infrared laser scanner [9]. By moving the robot along a given grid the laser moved over a surface. For each pose a high dynamic range image was taken and the features were extracted. Further, we extracted the skin thickness from MRI scans as a ground truth. Then we applied our setup to three subjects and could show that the scans correspond exceedingly well to our simulation results, allowing the detection of the skull with sub-millimeter accuracy [10], [11]. Nevertheless, a huge drawback of our approach was the scanning time which was approx. 45 min for the forehead. This is of course not practicable for clinical use and we must modify our optical setup. For this purpose we added a scanning device to our original setup. Now the laser is scanned by a galvanometric scanner and not by the robot. Here, we estimate the influences of this modification on our acquired data by performing scanning simulations.

2 Material and Methods

To improve the sampling rate of the optical setup, either the robot can be moved faster or the laser beam has to be moved by some other means. For tracking in real time, a grid resolution of 20 by 20 points ten times per second - 4k points per second - is necessary to reliably track objects. Because this is impossible with a robot, we decided to add gal-

vanometric scanning mirrors to our initial setup, which is described in more detail in [9]. The setup is aligned such that the optical axis of the high dynamic range camera (Andor Zyla) is parallel to the laser beam. A beamsplitter is used to overlay both axes. In addition, the optical axis of the camera is centered with respect to the scanning mirrors. Therefore, it is possible to look along the laser beam (*inBeam* or *beam's eye view*) during the whole scanning process. Our simulations showed that this configuration prevents disturbances produced by an angle between the laser- and camera axes.
Figure 1a shows a typical image of one point on the forehead captured with this setup. Similar to our simulations, we extract the optical features and estimate the skin thickness for every point on the forehead. In contrast to the robot setup, the angle between the surface normal of a point on the forehead and the laser beam increases significantly. We know from our simulations that reliable estimation of soft tissue thickness is possible for angles up to 45°. Therefore, the area A_F, i.e. the number of points on the forehead where the incident angles are smaller than 45° (Fig. 2 marked in black), is decreased. Note that it is not possible to refocus the sample because the optical setup is fixed at a given distance. Consequently, the depth of field of the high dynamic range camera has to be higher than the vertical expansion of the forehead. Adapting the size of circles of confusion to the pixel resolution (1 pix = 6.5 µm) and the used lenses (Computar M7528-MP, f = 75 mm), the depth of field can be determined.

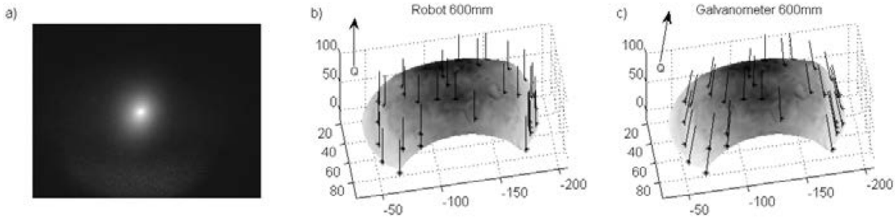

Figure 1: a) typical *inBeam* image captured with the optical setup, b) laser vectors (black lines) for robot setup on forehead (gray), c) laser vectors (black lines) for galvanometer setup

However, it is necessary to simulate the scanning process to estimate the expected angles and the area A_F to realize reliable measurements. For this purpose, we rotated forehead pointclouds extracted from MRI scans (resolution of 0.15 × 0.15 × 1 mm³) such that the *x*-*y*-plane was parallel to the face. Next, the central point on the forehead in *x*-*y*-direction was detected and its *z*-value was offset by the measuring distance *d*, the distance between our optical setup and the forehead. For our robot measurements, we had kept *d* at 300 mm [9]. Afterwards we determined the surface normals for each point of the forehead. Then we determined the vector between the laser source *Q* (Fig. 1b-c, black arrow in upper left corner) and the points of the forehead (*laser vector*, Fig. 1b, black lines). Finally, we computed the angle between the laser vectors and the surface normals. Additionally, we determined the needed depth of field (*nDOF*) for the given measuring distance *d* by subtracting the longest and the shortest laser vectors (which is valid due to our *inBeam*-configuration).
For comparison, the same simulation was performed for the robot setup. In contrast to the scanning mirror setup, the *x*- and *y*-components of *Q* were the same as the values of the point on the forehead. Additionally, *d* was kept constant, i.e. the *z*-value of *Q* was the *z*-value of the point offset by *d*. The results are shown in Fig. 1c.

3 Results

Figure 2a shows the simulated scanning process with the robot setup. Figure 2b shows the same for the scanning mirror setup for different measuring distances *d* ranging from 300 mm to 600 mm. As expected, the angular distribution for the forehead changed. Black lines in the plot encompass the area A_F where the incident angles are smaller than 45°.
During the measurements with the robot setup, angles of up to 45° are located in the outer region of the forehead. These angles are present closer to the center for the galvanometer setup (shift of the black line). The black pixels represent an incident angle of 45° ± 0.1°. It was the same for the other two subjects and of course the area A_F is much smaller using a galvanometer. Note that the maximum incident angles are higher than 90° because the point clouds are extracted from MRI scans. The expansion of the MRI scans is larger than the volume which can be scanned with the optical setup. An angle higher than 90° would result in obstruction of the laser.
In addition we determined the *nDOF* for three subjects at four measuring distances from 300 mm to 600 mm. Table 1 gives an overview of the estimated *nDOF*s.

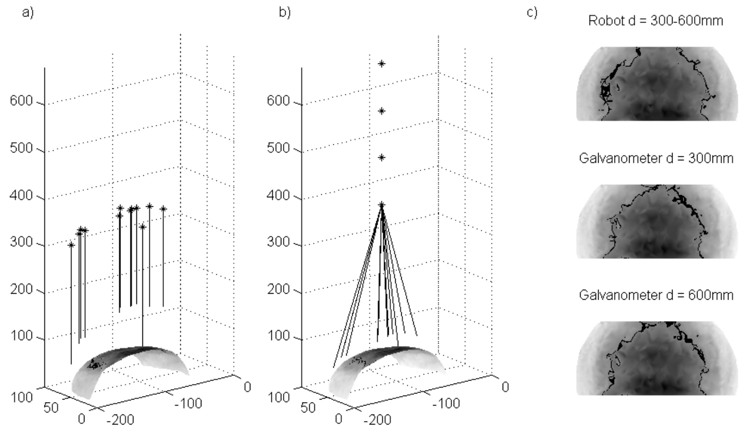

Figure 2: a) scanning process for robot setup with laser vectors (black lines), b) scanning process for galvanometer setup with laser source *Q* (black asterisks), c) angular distribution for robot and galvanometer setup for different distances *d* from 300 mm to 600 mm

d [mm]	$nDOF_{\text{subject 1}}$ [mm]	$nDOF_{\text{subject 2}}$ [mm]	$nDOF_{\text{subject 3}}$ [mm]
300	86.95	71.27	38.70
400	85.50	69.67	37.51
500	84.50	68.61	36.76
600	83.80	67.87	36.24
Δd [mm]	3.15	3.40	2.46

Table 1: Needed depth of field (*nDOF*) for different measuring distances *d* and subjects

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For the galvanometer setup, the needed *nDOF* varies between 2.46 mm and 3.4 mm. Here, higher measuring distances *d* result in lower values for the *nDOF*. For the robot setup, the *nDOF* is not important due to the measuring configuration. Note that for this determination, the absolute value of the *nDOF* depends on the extracted data from the MRI scans. For example subject 1 has a very long and high forehead whereas subject 3 has a very small forehead.

4 Discussion

We expected that the angular distribution changes when we use a galvanometer. This results in a smaller area A_F which is usable for skin thickness estimations to improve optical tracking. As a consequence of these results we have to adapt

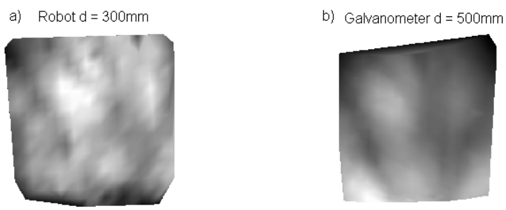

Figure 3: a) optical scan of a forehead using the robot setup at d = 300 mm, b): optical scan of a forehead using the galvanometer set-up at d = 500 mm.

our scanning grid resolution to the smaller area A_F. The area A_F is maximized when the measuring distance *d* is maximized. At the same time the *nDOF* decreases.

To realize a robust optical tracking system which can handle the problem of soft tissue artifacts, we must increase the measuring distance up to 600 mm. This would be the best tradeoff between resolution of the camera and depth of field.

Further, it is now possible to scan non-rigid objects much faster without using a robot and performing several calibration steps as it was necessary in [9]. This will substantially improve the system's accuracy
Nevertheless, real measurements show strong angle dependencies for the galvanometric setup. Figure 3a shows an optical scan of a forehead using the robot setup. Figure 3b shows the same forehead scanned using a prototypic galvanometric setup. As shown by our simulations [8] we assume that the blur is mainly caused by higher incident angles in that area.

5 Conclusion

We demonstrated a theoretical estimation for fast optical scans to improve optical tracking. We showed results of simulations as they could influence the acquired data. The *nDOF* decreases for higher measuring distances d whereby the number of points in the area A_F is maximized. As a consequence of these results, d should be maximized as long as it is possible to reliably extract optical features from the high dynamic range camera images. At the same time focus problems are handled. How far these angular influences are relevant for the results after post-processing the acquired data has to be evaluated in future work.

6 Acknowledgement

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7 References

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