

# Construction and testing of a computer-based intraoral laser scanner for determining tooth positions

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## Abstract

An optical set-up for intraoral data acquisition based on the principle of laser triangulation was developed. The system consists of a pig-tailed laser with line generating optics, a stepping motor driven positioning stage, a commercial CCD (charge coupled device) camera system with frame grabber interface, a control personal computer and a mirror system compensating for the fact that there is no possibility of watching an object directly in the mouth under a certain angle except from a facial position during intraoral scanning. Due to the size of the prototype measurements were still restricted to plaster casts. In order to evaluate its accuracy, the measurements were compared with those taken with a commercial laser scanner and a coordinate measurement table. The accuracy of the prototype scanner was determined to be  $\Delta XYZ=0.04$  mm using gauge blocks of given dimensions and proved to range between the commercial laser scanner and the coordinate measurement table (i.e., it was slightly better than that of the commercial scanner). Applications in orthodontics were demonstrated by scanning plaster casts and measuring distances on reconstructed surfaces. The measured distances showed a maximum deviation of about  $\pm 0.2$  mm compared with the data of the coordinate measurement table, which served as a reference. In addition, reconstruction of three-dimensional tooth movements was performed on the scan data. The translational and rotational parameters gained from the superimposition of scanned point clouds and describing tooth movement were also in good accordance with the reference. The achieved accuracy proved to be sufficient for further development which should include a reduction in size and the use of more precise device components. © 2001 IPPEM. Published by Elsevier Science Ltd. All rights reserved.

**Keywords:** Laser scanning; Intraoral; Triangulation method; Three-dimensional measurement; Tooth movement

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## 1. Introduction

In orthodontics an accurate and easy to perform method for retrieving three-dimensional data of tooth positions and orientations is of highest interest. The strain on the patient should be as small as possible and the measuring time and the space required for storage of the measuring aids should be minimized. Coordinates of tooth positions, if not manually measured directly in the mouth, are usually retrieved out of plaster casts. Several optical and mechanical methods have been developed for the analysis of plaster casts. For example, a holographical method has been put into practice as an

alternative to the space consuming storage of plaster casts [1,2]; the analysis of tooth positions, however, is lavish and faulty and the set-up is technically sophisticated. A very accurate method of measuring dental casts offers a coordinate measurement table, equipped with standardized linear encoders and suited to the needs of orthodontic applications [3]. Using this mechanical instrument, each point on the surface of a dental cast has to be found separately by a contact method. Thus the data acquisition rate is very slow (about 1 point per second). On the other hand its accuracy is 0.014 mm and its resolution amounts to 0.004 mm.

The recently presented Bending Art System (BAS [4]), which can be considered as an intraoral data-acquisition method, has been put into practice for the automated bending of dental arch wires. Using this system, specially designed marker elements have to be inserted into the slots of the orthodontic brackets serving as refer-

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ence points. With this method only the coordinates of bracket positions and not of the tooth positions can be determined. These have to be derived from the bracket system used.

Laser scanners are widely used in orthodontics and maxillo-facial surgery to achieve three-dimensional digitization of the jaw and face, respectively [5–12]. They represent an efficient non-contacting optical method to capture large amounts of data in a computer within a short time, as the acquisition rate for laser scanners is about 5.000 datapoints per second. Direct three-dimensional digitization of the shape of a jaw into a computer allows the immediate access on the data and the instant execution of numerical calculations, for example the surface reconstruction or the determination of three-dimensional tooth movements. Up to now, laser scanning was restricted to the scanning of plaster casts, since the size of the devices is not yet suitable for an intraoral application.

It was the aim of this study to develop a device that combines the advantages of a three-dimensional laser scanner with an intraoral acquisition method. The different components that form the complete system are basically commercially available electronic and optical devices, respectively. The components and the set-up are described in the following section.

## 2. Components of the set-up

Fig. 1 shows a schematic diagram of the whole system. It consists of the following components:

- laser diode with fibre pigtail and line-generating optics;
- two flat mirrors;
- stepping motor driving the set-up consisting of the laser and the mirrors;
- stepping motor booster to generate the pulses for the stepping motor;

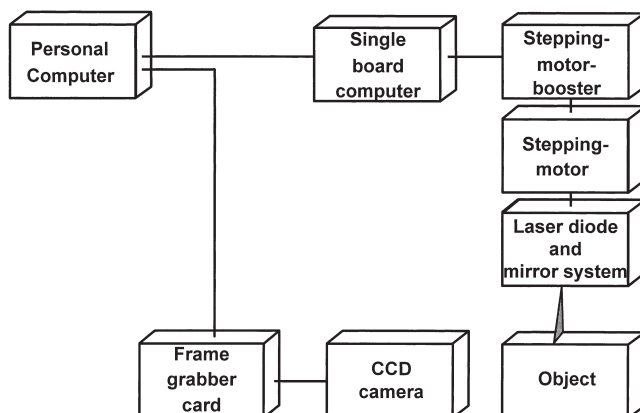


Fig. 1. Schematic diagramme of the developed laser scanning system.

- single-board computer running the control software for the stepping motor;
- CCD (charge coupled device) camera and frame grabber board; and
- a personal computer running the control software.

### 2.1. Optoelectronic subsystem

A laser diode (690 nm, POS-PLD-50, Laser 2000, Germany) with a fibre pigtail and line-generating optics at its end produces a laser line illuminating the object to be scanned, making contour lines on its surface visible. A CCD camera views at this scene to measure the illuminated profile under a certain angle (the triangulation angle). Fig. 2 illustrates the arrangement of the two mirror system, the CCD (charge coupled device) camera and the fibre optics, which takes into account the space problems involved in fitting the device into the mouth and simultaneously considering the triangulation angle. The use of two mirrors makes it possible to look at the object from two different views under different angles: a facial as well as a dorsal view. The two data sets can be obtained with one measurement process.

This can be seen in Fig. 3, which shows the assembly of the prototype. The CCD (charge coupled device) camera (ST-30H, Stemmer, Germany) is placed at a distance of about 400 mm in front of the mirrors. It is connected to a frame grabber card (ITI IC-P-4M, Stemmer) that takes over the video data and feeds them into the memory of a personal computer (PentiumPro series computers minimum, running under Windows NT with a minimum of 128 MB RAM). Due to the inclination of the mirrors the camera sees the illuminated profiles in the same way, as if it was positioned under a certain triangulation angle towards the horizontal, which in this case corresponds to twice the inclination angle  $\alpha$  of the mirrors seen in Fig. 2.

The measuring range of the device is 70 mm in depth (X-axis), which corresponds to the mean depth of the focus of the camera and its lens system (TV lens B 5014A-TH, Stemmer,  $f=25$  mm, 1:1.4). The measuring range along the Z-axis is limited by the inclination and size of the larger and the smaller mirror and is 55 mm and 10 mm in height, respectively. The width of the mirrors do not exceed 60 mm, in order to represent an appropriate size for a standard adult and thus limiting the measuring range in the Y-direction.

### 2.2. Electronic and mechanical subsystems

Scanning of the complete surface of an object with a laser scanner requires a transversal shift of the object underneath the laser line. Thus, subsequent contour lines can be registered, representing the three-dimensional surface by a cloud of points. For the future intraoral

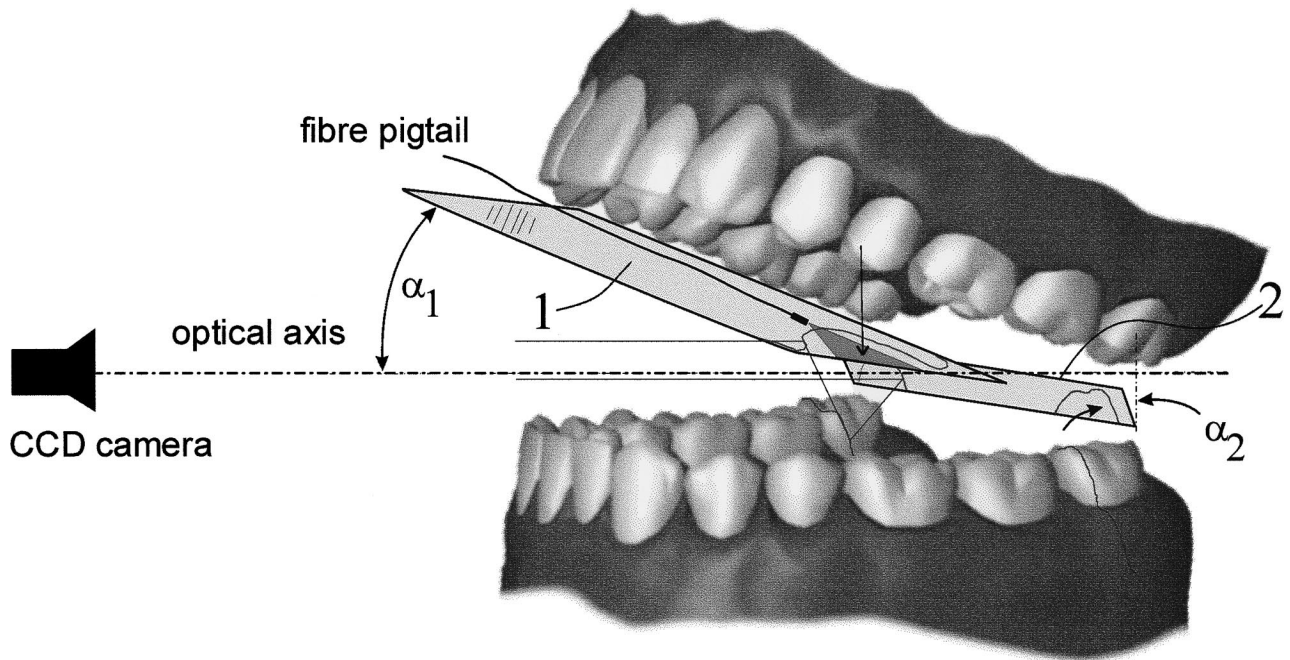


Fig. 2. Principal arrangement of the mirrors (1 and 2), the laser diode and the CCD camera of an intraoral laser scanner which fits into the patient's mouth.

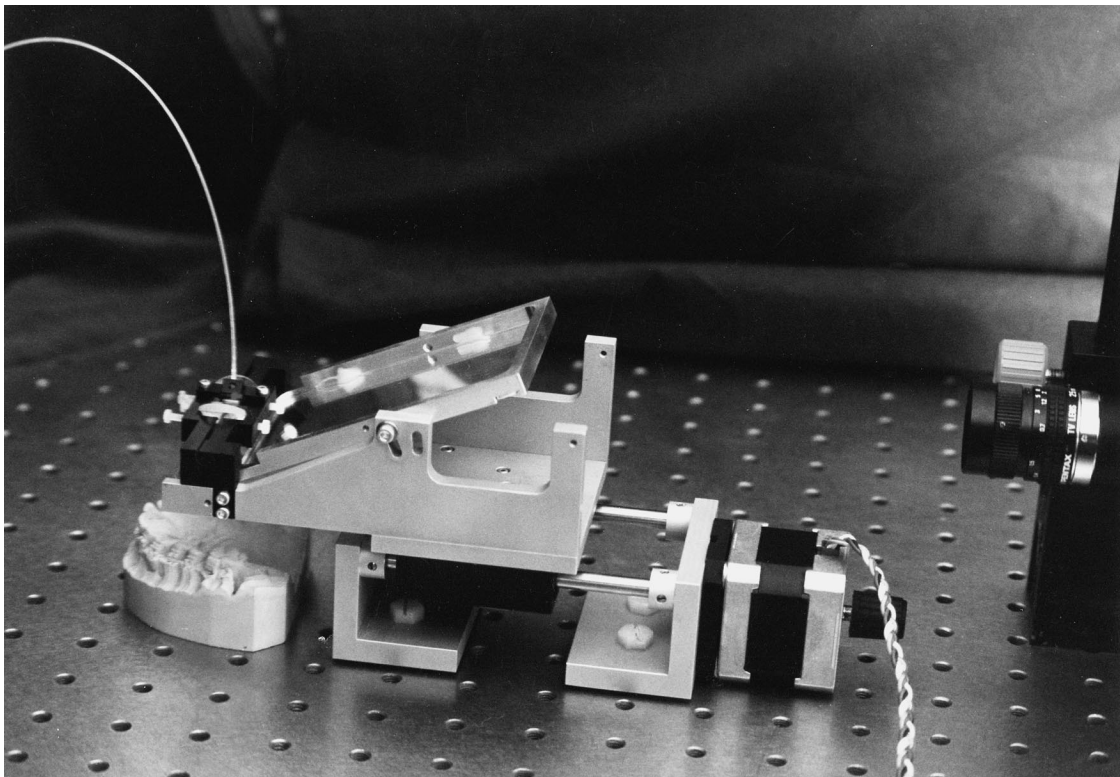


Fig. 3. The assembly of the prototype, which demonstrates the principle of intraoral scanning. On the left the assembly of the stepping motor driven translatable stage with mirror system and fibre pig tail scanning a plaster cast can be seen. On the right there is the CCD camera with the lens system.

application of our laser scanner, the measuring object, that is, the patients' jaws, cannot be moved. Thus, a stepping motor driven positioning stage was designed moving the system of laser and mirrors along the optical axis. By these means subsequent contour lines are extracted out of CCD pictures. A single board computer (SAB 80515, Siemens, Germany) is used as an interface between the stepping motor driver (SINCOS-L, Phytron, Germany) and the personal computer. This interface computer receives the control signals from the personal computer, advising the stepping motor to change position. Acceleration and deceleration ramps for the stepping motor booster as well as the adequate number of stepping pulses for a certain measuring range are generated by the single board computer autonomously.

The control software of the personal computer was developed to maintain an organized measurement sequence without timing problems: two independent processes (so-called threads) are running simultaneously during data acquisition. They include acquisition of the video signals by the frame grabber card and a steady data flow to and from the single-board computer. Further tasks of the control software, following data acquisition, include calculations of real three-dimensional space coordinates out of screen coordinates and rendering of the scanned object surfaces represented by the point clouds [13,14].

### 2.3. Image reproduction

The relationship between the screen coordinates and the object coordinates in real three-dimensional space can be derived using several relationships from basic geometrical optics and all relevant distances and angles in the set-up [13]. Actually, three device coordinates are necessary to calculate the corresponding coordinates in real space, namely the current position of the stepping motor derived from the number of steps  $n$  and the screen coordinates  $B_Y$  and  $B_Z$  delivered by the frame grabber card. The  $X$ -coordinate can directly be calculated from  $n$ , as the coordinate system is chosen in a way that the  $X$ -axis points into the direction of movement of the stepping motor and coincides with the optical axis of the device.

The use of two mirrors results in two independently generated contour lines. As can be seen from Fig. 4, showing the two different views of a certain scan line, the arrangement of the mirrors guarantees that there are no intersections between the two lines. Thus, extraction of the screen coordinates from a CCD picture is simply performed from each half of the screen. Line extraction takes 15 ms for one camera shot. Triangulation of the screen coordinates to calculate the real three-dimensional coordinates is now performed with the two lines using the triangulation angle  $\gamma_i = 2\alpha_i$ , where  $\alpha_i$  is the inclination angle  $\alpha_{1/2}$  of the corresponding mirror. The tri-

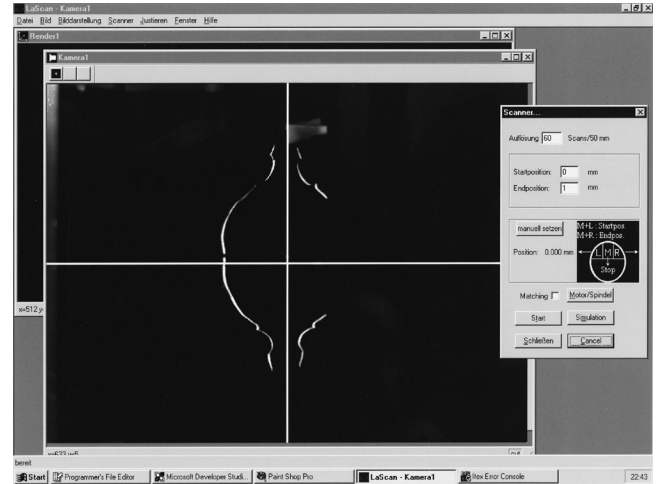


Fig. 4. Screen dump of the software with the two different views seen by the CCD camera.

angulation algorithm, applied, compensates for the change in triangulation distance due to the movement of the mirror system and takes into account the true size of the CCD chip that differed from the dimension stated by the manufacturer. In total, the following triangulation relationships could be derived for the calculation of real three-dimensional coordinates from the views of the two mirrors [13]:

$$\vec{X}_{\text{mirror1}}(n, B_Y, B_Z) = \begin{pmatrix} -n \frac{S}{8000} \\ B_Y \frac{d_Y^{\text{CCD}}}{572} \left( \frac{g(n)^{\text{mirror1}}}{f} - 1 \right) \\ B_Z \frac{d_Z^{\text{CCD}}}{768} \left( \frac{g(n)^{\text{mirror1}}}{f} - 1 \right) / \cos(2\alpha_1) \end{pmatrix},$$

$$\vec{X}_{\text{mirror2}}(n, B_Y, B_Z) = \begin{pmatrix} -n \frac{S}{8000} \\ B_Y \frac{d_Y^{\text{CCD}}}{572} \left( \frac{g(n)^{\text{mirror2}}}{f} - 1 \right) \\ B_Z \frac{d_Z^{\text{CCD}}}{768} \left( \frac{g(n)^{\text{mirror2}}}{f} - 1 \right) (1 - \tan(\alpha_2) \tan(2\alpha_2)) \end{pmatrix},$$

where  $\vec{X}_{\text{mirror1,2}}$  are the real three-dimensional coordinates of an individual point on a scan line of mirrors 1 and 2, respectively, and  $n$  is the number of steps of the stepping motor. With the pitch  $S$  of the spindle (0.5 mm per revolution) and the number of 8000 impulses per full  $2\pi$  rotation of the motor, the  $X$ -coordinate may be calculated. The  $Y$ - and  $Z$ -coordinates are dependent on the measured picture sizes in pixel  $B_{Y,Z}$  on the CCD chip, the chip dimensions  $d_{Y,Z}$ , the inclination angles  $\alpha_{1,2}$  of the corresponding mirror and the focal length  $f$  of the lens system. The object distance  $g^{\text{mirror1,2}}$  is dependent on the distance of the mirrors from the lens and

thus dependent on the number  $n$  of steps of the stepping motor.

Subsequently, the two lines have to be combined into a continuous one. A matching algorithm was developed, that basically consists of a minimization of the distance of individual points describing the course of the two lines: the lines can be expected to be as close together as possible, if the sum of the distances from each point of the first line to each point of the second line results in a minimum value. Approximation of the course of the lines is achieved by applying a linear transformation (translation in  $Y$  and  $Z$ , rotation around  $X$ ) to the points of line 2 until the distance is minimized. For those regions of the profile where two points are very close together and have a common  $Y$ -value, the mean value is calculated.

#### 2.4. Performance of the set-up

Point clouds with different numbers of scan lines are shown in Fig. 5. As density rises [from (a) through (c): 140, 320 and 430 scanlines], details of the surface become more and more visible. Even an arrow, written with dark felt pen on the cast (see right side of each picture), could be made visible. The figures give an impression of the acquisition range of the device. It can be seen from the pictures that dark spots exist in the point clouds. These correspond to parts of the cast which cannot be seen by the camera even if they are illuminated. This can be attributed to the triangulation principle realized here: the laser light seen by the camera is emitted under an angle of twice the inclination angle  $\alpha_i$  of its corresponding mirror. If parts of the surface are steeper than that they prevent a reflection by simply covering the light. This especially holds for the palatal surfaces of the incisor teeth.

The time required for generating a point cloud of the size seen in Fig. 5c was about 2 min. A total of 1 minute was required for data acquisition, including the pure acquisition time of the frame grabber card as well as the time necessary for moving the stepping motor and performing line extraction out of the subsequent CCD pictures. Minimizing the positioning time of the stepping motor could result in a total acquisition time as short as 6 s. About 1 min is spent on performing the matching of the two different views. This may be reduced essentially with the use of a high performance computer.

#### 2.5. Adjustment, resolution and accuracy

Adjustment of the set-up was performed in several steps. First of all the camera had to be aligned exactly along the defined optical axis, as shown in Fig. 2. In the next step, all relevant lengths and angles had to be determined, either directly by measuring or indirectly, as in the case of the triangulation angles, that is, the

angles of the mirrors. Their values could be derived from measurements of objects with exactly known dimensions with the completed set-up and by back-calculation (gauge blocks with  $L=2.0\text{ mm}\pm 0.2\text{ }\mu\text{m}$ ,  $10.0\text{ mm}\pm 0.3\text{ }\mu\text{m}$  and  $20.0\text{ mm}\pm 0.3\text{ }\mu\text{m}$ , accuracy after DIN 861: 'Gütegrad 1', Hommel-Hercules Werkzeughandel GmbH and Co. KG, Viernheim, Deutschland). In the case of reproduction of the correct value, the corresponding angle could be expected to be accurate. This procedure was performed independently for each mirror.

To ensure that all parameters were determined with sufficient accuracy, calibration measurements of gauge blocks in horizontal and vertical alignments were performed. The results are displayed in Fig. 6. It can be seen that there was a maximum deviation of about  $\pm 0.2\text{ mm}$  around a rated value. The mean variance of the data sets could be determined to be  $0.04\text{ mm}$ . In Fig. 6 similar measurements performed with a commercial laser scanner (Micromasure 70®, microdenta, Linden, Germany) are included, showing maximum deviations of the same size. A variance of about  $0.05\text{ mm}$  was determined for this device [15].

The resolution of the developed system was determined to be 3180 (human eye: 4926), which is equivalent to a height of  $0.3\text{ mm}$  in a distance of  $1\text{ m}$  [13]. This means that the smallest element into which the screen picture may be divided is 1 pixel. Thus, the limits of the resolution and accuracy are caused by the different components of the basic principle of a laser scanning device: the resolution of the CCD camera; the width of the laser line, which has a finite expansion; and laser noise. The CCD chip used in this study was  $768\times 572$  pixel in its dimensions. The use of a higher resolution thus is essential for improving accuracy.

### 3. Overall test of the apparatus — results

#### 3.1. Measurement of distances

To test the accuracy of the set-up, measurements of reference points on plaster casts were performed. The reference points were measured with a coordinate measurement table as well, and the results served for comparison and reference, as the coordinate measurement table has a high degree of accuracy [3]. Fig. 7 depicts the surface of an especially designed plaster cast with tetrahedronal imprints on its surface. The tips of the tetrahedrons served as reference points and the distances between selected points were determined. For better visibility a surface was generated from the point cloud using triangulation algorithms.

The maximum deviation of the measurements with the laser scanner from the values obtained with the coordinate measurement table was about  $\pm 0.2\text{ mm}$  (see Table 1, last column). It must be noted that this value rep-

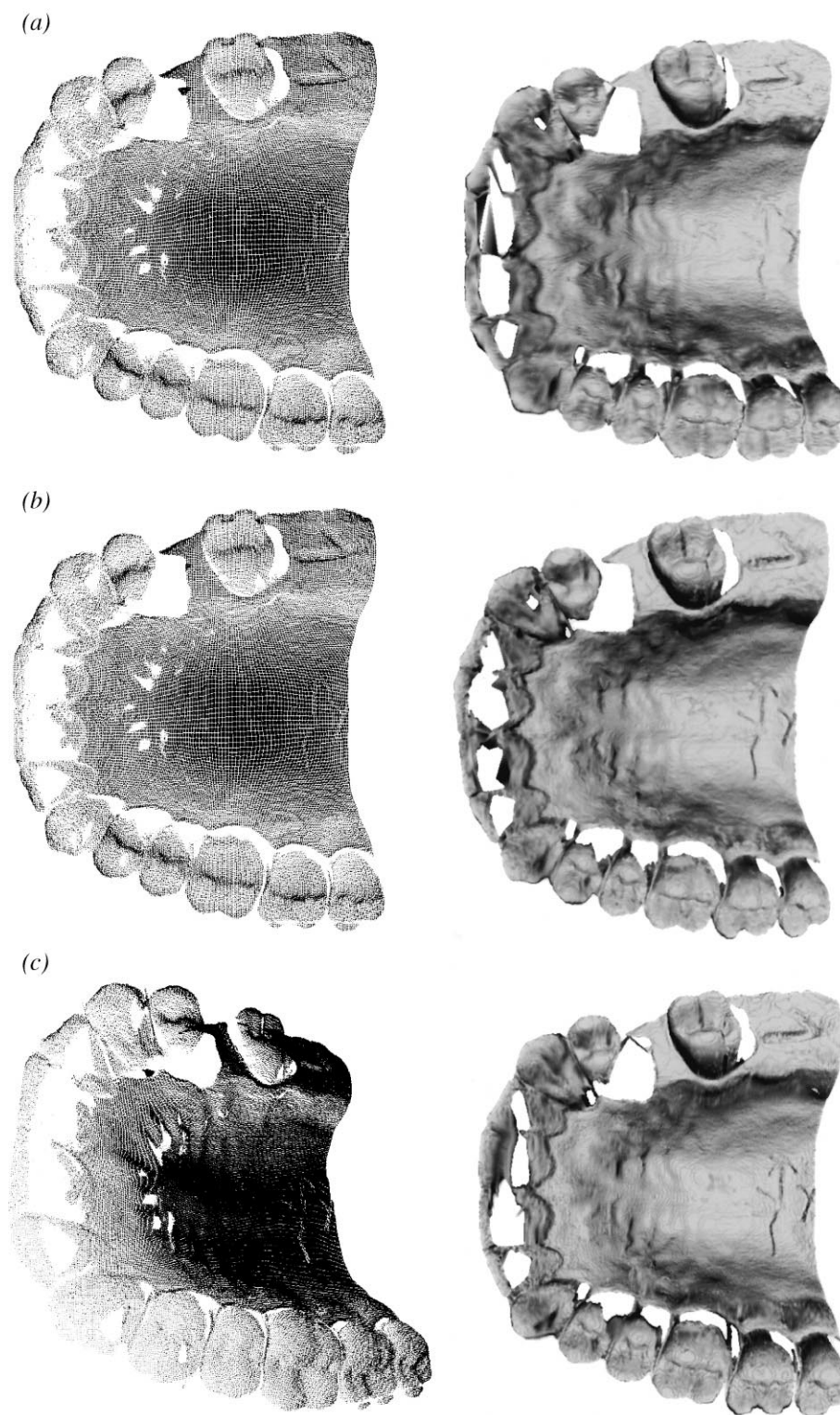


Fig. 5. Point clouds with different numbers of scanlines. A surface generating algorithm, especially adapted to the format of the scan data, has been developed and produces a surface within seconds. Imperfect reproduction of the surfaces results from large areas in the point clouds that could not be scanned due to shadow effects.

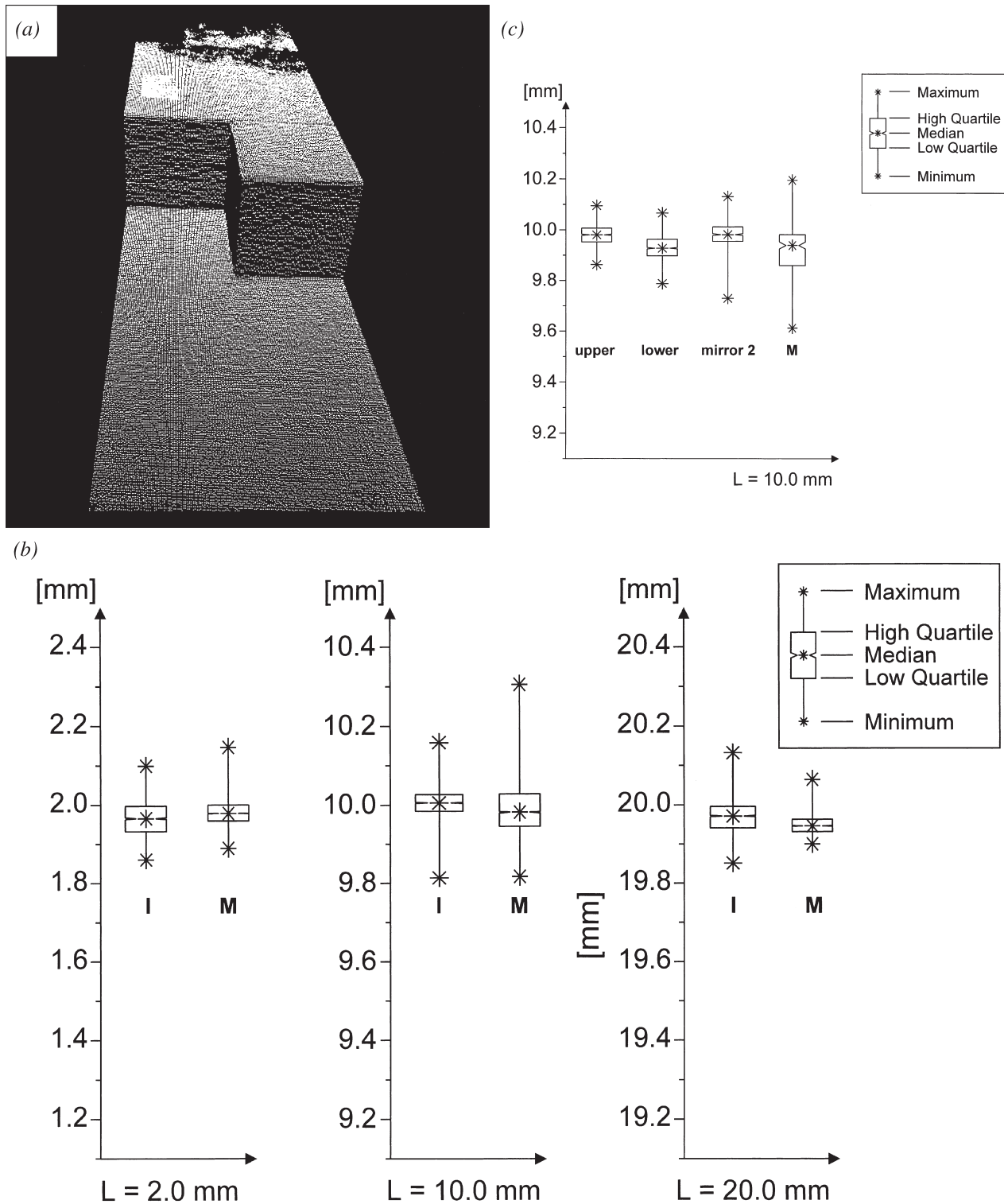


Fig. 6. Measurement of gauge blocks of different dimensions with our intraoral laser scanner (I) and with the commercial scanner, Micrometre 70® (M). (a) Three-dimensional view of a point cloud of two gauge blocks arranged to calibrate the Y-axis as measured with our scanner. (b) Calibration of the Z-direction. The deviation of about  $\pm 0.2$  mm from the rated value is due to the resolution of the CCD chip, the extension of the laser line and laser noise. (c) During calibration of the Y-axis, a slight dependence of the measured dimension of the gauge block from the position under mirror 1 was determined. This is indicated by the two positions 'upper' and 'lower' for the measurements of the 10 mm gauge block. There was no dependence of the dimension as measured by mirror 2 exclusively. The variance of the values determined using the commercial scanner (M) is slightly larger.

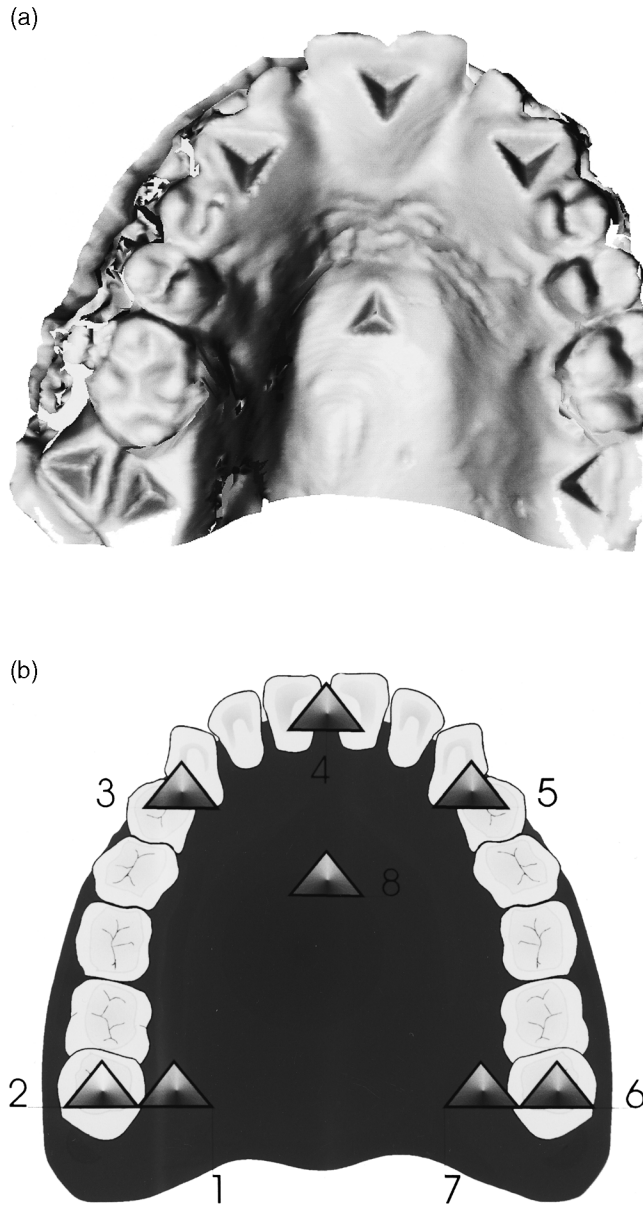


Fig. 7. An especially designed plaster cast that allows measurements of distances. The tips of the tetrahedrons serve as reference points. (a) Reconstructed surface of the scanned point cloud, (b) identification of the reference points in the cast.

resents the mean value of multiple measurements. One single measured distance on a point to point basis may have a deviation of up to 0.6 mm, directly being the consequence of the limited resolution and accuracy of the laser scanning principle.

### 3.2. Reconstruction of orthodontic tooth movements

The benefit of a laser scanner is the large number of data points defining the surface of an object, registered within a limited time interval. Thus, the orthodontic application of such a device should take advantage of

this large number of points since the surface of an object defines its position and orientation. Orthodontic tooth movements consist of three-dimensional translations and rotations. For the reconstruction of an orthodontic tooth movement, jaws or plaster casts of intraoral situations before and after a certain treatment step have to be measured: the first point cloud represents the initial situation whereas the second one represents a final situation, where teeth have been moved and/or rotated. The translational ( $X, Y, Z$ ) and the rotational ( $\phi, \theta, \psi$ ) parameters of the movement can be determined by a numerical superimposition of the two point clouds. This is equivalent to the minimization of a function  $f(X, Y, Z, \phi, \theta, \psi)$  that defines the distance between the point clouds by the sum of the distances of each individual point in cloud one and two. The process is called surface-surface-matching [16] and was implemented using an iterative procedure.

The principle was tested with a set of especially prepared plaster casts. The first cast serves as a reference, defining the initial treatment situation, in the second one molar tooth was taken out and repositioned at a given distance to the original position. Both casts were scanned and additionally measured with the coordinate measurement table. Fig. 8 shows the subsequent stages of the superimposition process with data sets taken by our intraoral laser scanner. Surfaces have been generated from the point clouds and the data sets have been segmented into separate clouds of the palate, the moved molar and two further molars, serving as control teeth for the test of the accuracy of the whole matching process in conjunction with data taken either by the laser scanner or by the coordinate measurement table.

Determination of the tooth movements was done in the following steps:

1. The surface of the palate of cast 2 was matched to the surface of the reference cast 1, giving a set of three translations and three rotations. Fig. 8a shows both plaster casts in different coordinate systems, due to subsequent scanning with different positions in the scanner.
2. A transformation using these parameters was applied to the teeth of cast 2. Now, palates and teeth of both casts were in a common reference frame. The result of moving the surfaces of the second cast into the coordinate system of the first one is shown in Fig. 8b. The movement of the right molar can be recognized by the offset of the lighter molar surface with respect to the darker one and more or less stable positions of the reference teeth on the left side.
3. Final matching procedures of all transformed teeth of cast 2 to the corresponding teeth of cast 1 deliver the movement parameters ( $X, Y, Z, \phi, \theta, \psi$ ) from the function minimization for each individual molar. The result is shown in Fig. 8c. The movement parameters

Table 1

Comparison of the measured distances with our laser scanner (IOLaScan) with those determined using a coordinate measurement table (CoordMeasT) — for definition of the straight lines see Fig. 7

| Distance | IOLaScan (mm) | SD    | CoordMeasT (mm) | SD    | Deviation (mm) |
|----------|---------------|-------|-----------------|-------|----------------|
| P2P1     | 6.692         | 0.094 | 6.752           | 0.008 | 0.060          |
| P2P3     | 34.207        | 0.116 | 34.175          | 0.011 | −0.035         |
| P2P4     | 47.524        | 0.075 | 47.555          | 0.009 | 0.031          |
| P2P5     | 52.510        | 0.122 | 52.464          | 0.006 | −0.046         |
| P2P6     | 56.620        | 0.114 | 56.389          | 0.013 | −0.231         |
| P5P3     | 27.081        | 0.021 | 26.881          | 0.003 | −0.200         |
| P5P4     | 15.156        | 0.088 | 15.180          | 0.006 | 0.024          |
| P5P6     | 33.154        | 0.109 | 33.080          | 0.007 | −0.074         |
| P6P7     | 7.448         | 0.096 | 7.561           | 0.014 | 0.113          |
| P8P1     | 33.060        | 0.088 | 33.029          | 0.011 | −0.031         |
| P8P4     | 29.282        | 0.075 | 29.330          | 0.009 | 0.048          |
| P8P6     | 36.806        | 0.094 | 36.886          | 0.014 | 0.080          |

Table 2

Comparison of the results of the reconstructed tooth movement for the intraoral laser scanner and the coordinate measurement table

|              |            | Translation (mm) |          |          | Rotation (°) |          |        |
|--------------|------------|------------------|----------|----------|--------------|----------|--------|
|              |            | <i>X</i>         | <i>Y</i> | <i>Z</i> | $\phi$       | $\theta$ | $\psi$ |
| Molar        | CoordMeasT | −5.83            | −0.99    | −2.23    | 2.52         | 1.02     | 0.22   |
|              | IOLaScan   | −6.30            | −0.96    | −2.46    | 2.20         | 0.76     | 0.27   |
| Ref. tooth 1 | CoordMeasT | −0.16            | 0.07     | −0.21    | 0.67         | −0.70    | 0.66   |
|              | IOLaScan   | 0.25             | 0.08     | −0.13    | 0.23         | −0.76    | 0.01   |
| Ref. tooth 2 | CoordMeasT | −0.11            | 0.11     | −0.39    | 0.03         | −0.01    | 0.01   |
|              | IOLaScan   | 0.43             | 0.04     | −0.28    | 0.32         | −1.60    | 0.16   |

Table 3

Assessment of the accuracy of the movement reconstruction using the reference teeth; the mean, the standard deviation and the maximum deviation have been calculated from the values in Table 2

|            | Error in translation (mm) |          |                   | Error in rotation (°) |          |                   |
|------------|---------------------------|----------|-------------------|-----------------------|----------|-------------------|
|            | Mean value                | Variance | Maximum deviation | Mean value            | Variance | Maximum deviation |
| CoordMeasT | 0.18                      | 0.11     | 0.43              | 0.42                  | 0.32     | 1.01              |
| IoLaScan   | 0.17                      | 0.13     | 0.43              | 0.56                  | 0.46     | 1.60              |

(*X*, *Y*, *Z*,  $\phi$ ,  $\theta$ ,  $\psi$ ) are given in Eulerian angles and in translations with respect to the origin of the coordinate system of plaster cast 1. For an application in orthodontics these may be transformed into coordinate systems with respect to each individual tooth.

Basically all transformational parameters of the reference teeth should turn out to be zero in the course of the minimization process. Any deviation from zero should therefore represent an appropriate measure for estimating the error of the whole procedure. Orthodontic tooth movements range up to 5 mm for translations and several 10 degrees for rotations, respectively. An error of up to 10% should be adequate. In Table 2 the results of the

movement reconstructions using either the data taken with the laser scanner or the data taken with the coordinate measurement table are listed.

It is obvious that the results for the moved molar delivered by the two different experimental methods differ by not more than  $\pm 0.5$  mm in translations and  $\pm 0.5^\circ$  in rotations. The same holds for the reconstructed movements of the reference teeth. Calculating mean values out of all the deviations from zero for the reference teeth delivers ca 0.18 mm in translations and  $0.5^\circ$  in rotations for both devices (Table 3). These results show that the errors resulting from determining tooth positions based on laser scanning data can be expected to range in the same order of magnitude as with the use of a coordinate

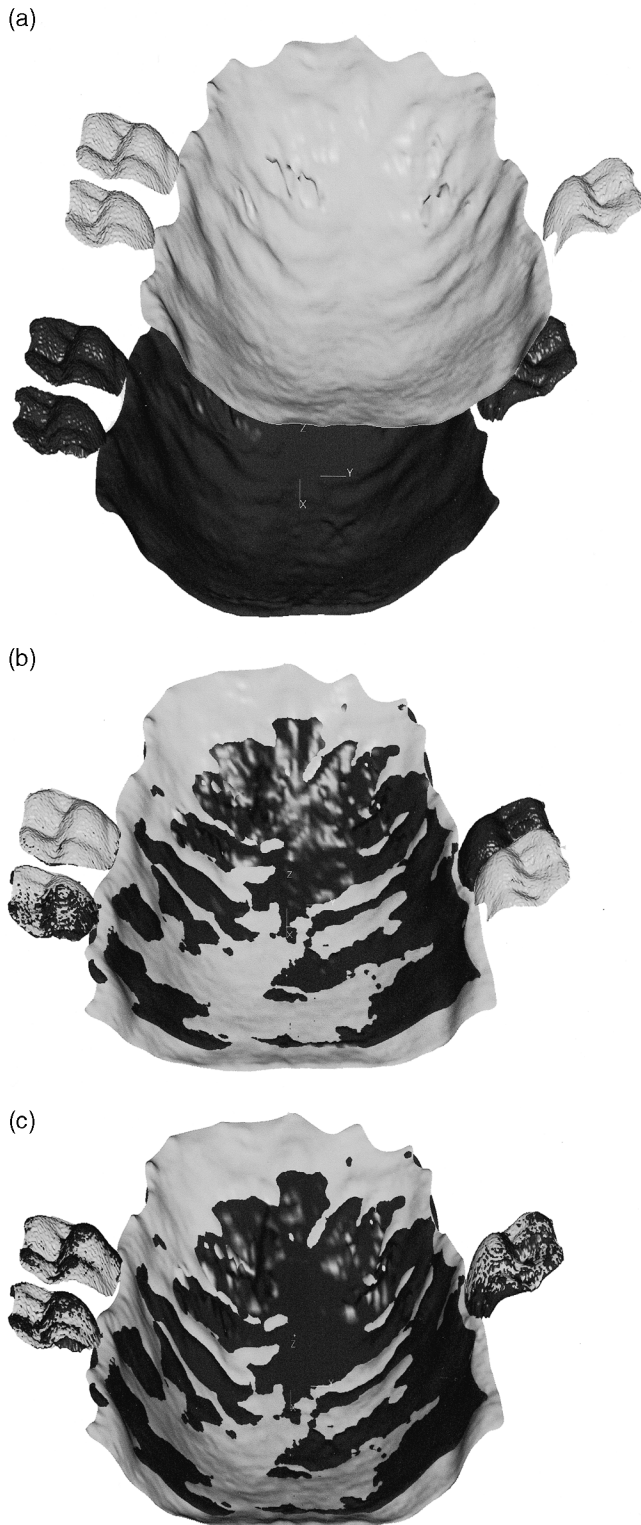


Fig. 8. Segmented point clouds with reconstructed surfaces as measured with our intraoral scanner during different stages of tooth movement. (a) Separated casts, (b) matching of the palate to create a common reference frame, (c) matching of the moved molar and the reference teeth.

measurement table, although the accuracy for single point measurements of the latter one is significantly higher.

#### 4. Conclusions

The optomechanical device described in this paper was developed to offer a data acquisition method for retrieving tooth positions and orientations, based on the principle of laser triangulation, directly in the mouth. The development of the prototype showed that intraoral measuring and three-dimensional digitization of tooth positions can be combined to one compact device. The time span necessary for data acquisition, that is, the time the apparatus has to stay in the patient's mouth, coincides roughly with the time an orthodontist needs to take an impression with standard methods (60 up to 120 s). Thus it can be concluded that the strain on the patient will not be increased significantly by the use of the intraoral laser scanner. The main challenge in designing the intraoral device was the problem of limited space for the mirrors and the laser diode with the line generating optics. These devices have to stay in the mouth during the measuring process.

The method of using two mirrors (according to the arrangement in Fig. 2) could not yet be tested in patients directly, as the apparatus needs to be reduced in size by ca 20% more. If the strain on the patients seem to be too high, a solution with only one mirror might be necessary, although there will be the disadvantage of a loss of data of ca 25%. Consequently multiple scans will have to be performed with different mirror orientations, resulting in increased technical specifications and increased scanning times.

The results on the accuracy of our laser scanner prototype seem to be very promising, except for point to point measurements. However, this is a basic problem of the measurement principle involved. There are various ways to improve the accuracy if reference points need to be measured:

- Use of a CCD-chip with significantly increased resolution.
- Increasing the laser intensity allows the use of a smaller diaphragm of the camera, resulting in greater sharpness of the picture. However, this has its limitations in patient usage.
- Decrease of the width of the laser line.

Although the developed device cannot and does not need to replace plaster casts at present, it represents an alternative to deliver three-dimensional data on tooth positions that can be used for a computer aided orthodontic treatment planning system. Future work will concentrate on the integration of intraoral acquisition methods and

software developments to support the orthodontist in treatment planning.

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