RESEARCH

A method for measuring three-dimensional mandibular kinematics *in vivo* using single-plane fluoroscopy

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Objectives: Accurate measurement of the three-dimensional (3D) motion of the mandible *in vivo* is essential for relevant clinical applications. Existing techniques are either of limited accuracy or require the use of transoral devices that interfere with jaw movements. This study aimed to develop further an existing method for measuring 3D, *in vivo* mandibular kinematics using single-plane fluoroscopy; to determine the accuracy of the method; and to demonstrate its clinical applicability *via* measurements on a healthy subject during opening/closing and chewing movements.

Methods: The proposed method was based on the registration of single-plane fluoroscopy images and 3D low-radiation cone beam CT data. It was validated using roentgen single-plane photogrammetric analysis at static positions and during opening/closing and chewing movements.

Results: The method was found to have measurement errors of 0.1 ± 0.9 mm for all translations and $0.2^\circ \pm 0.6^\circ$ for all rotations in static conditions, and of 1.0 ± 1.4 mm for all translations and $0.2^\circ \pm 0.7^\circ$ for all rotations in dynamic conditions.

Conclusions: The proposed method is considered an accurate method for quantifying the 3D mandibular motion *in vivo*. Without relying on transoral devices, the method has advantages over existing methods, especially in the assessment of patients with missing or unstable teeth, making it useful for the research and clinical assessment of the temporomandibular joint and chewing function.

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Introduction

Knowledge of the kinematics of the mandible relative to the maxilla is helpful for a better understanding of the normal function of the temporomandibular joint (TMJ), and for the study of the aetiology, diagnosis and subsequent treatment of temporomandibular disorders.^{1,2} It also has a profound influence on the development of articulators and on the evaluation of the health of the

masticatory system.^{3–5} Therefore, accurate measurement of the three-dimensional (3D) motion of the mandible relative to the maxilla is essential for relevant clinical applications.

3D mandibular movements were first recorded using a complicated and bulky device with two mechanical face bows.^{6,7} Since then, much more compact and delicate systems have been developed, making significant contributions to clinical dentistry.⁸ However, all these measuring devices have to be attached onto the head and/or the mandible, affecting the normal motion of the jaw. Non-contact measurement systems address this problem by using in-house-developed camera

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systems,⁹ commercially available optoelectronic tracking systems^{10–12} or accelerometers.¹³ Some of these methods used skin markers for motion tracking, but errors associated with skin movements relative to the underlying bone were inevitable.¹² Transoral measurement devices attached directly to the teeth have been used to remove skin movement artefacts, but these devices might interfere with the jaw movements. Therefore, there is a need for an accurate method for measuring 3D mandibular motion for clinical and research purposes without using devices that may impede jaw movements.

Medical imaging techniques provide the most direct way of measuring joint motion without interfering devices. Single-plane fluoroscopy has been used in the description of the sagittal kinematics of chewing motion.¹⁴ MRI has also been used in the study of the TMJ in the sagittal plane.¹⁵ Slice-to-volume registration methods have been developed for accurate 3D, intraoperative positioning of the body parts and surgical instruments using either fluoroscopy or cine MRI.¹⁶ Since the bones of a joint may move out of the imaging slice during non-planar movements such as chewing, the application of slice-to-volume registration methods to mandibular kinematics measurement is limited. Another limitation of an MR-based method is the interference of metal components such as dental restorative components. Registration methods using bi-planar fluoroscopy have been proposed for accurate measurement of joint kinematics in three dimensions,¹⁷ but the radiation dose involved may limit their clinical applications. To date, their applications for dental purposes have not been published. A roentgen single-plane photogrammetric analysis (RSPA) method based on radio-opaque markers was developed to measure knee joint motion with rotational and linear accuracies of 0.27° and 0.9 mm, respectively.¹⁸ While this method is considered accurate enough for measuring TMJ movements, its use is limited because the markers used may affect natural oral movement, and it is not feasible when teeth are missing or unstable. Nonetheless, the RSPA methods may be used to validate other methods. More recently, Tsai *et al*¹⁹ established a volumetric model-based 2D-to-3D registration method for measuring 3D, *in vivo* kinematics of the knee with single-plane fluoroscopy. The technique can be further developed for accurately measuring 3D mandibular motion without using external devices or markers.

The aims of this study were to develop further a 2D-to-3D registration method for measuring 3D, *in vivo* kinematics of the mandible with single-plane fluoroscopy; to determine experimentally the accuracy and precision of the method; and to demonstrate the clinical applicability of the method *via* measurements on a healthy subject during opening/closing and chewing movements. It was hoped that, after validation, the method would be helpful for providing accurate *in vivo* 3D kinematic data of the mandible required for research and clinical applications.

Materials and methods

Subject preparation

A female (age, 35 years; height, 157 cm; mass, 50 kg) without missing teeth, dental prostheses or TMJ disorders participated in the current study with informed written consent as approved by the Institutional Research Board. Her maxilla and mandible were each fitted with four radio-opaque crystal markers (5 mm in diameter) which were attached to the embrasures of the central incisors, canine and first premolar, and molars (Figure 1). The marker positions relative to the bones were determined using cone beam CT (CBCT) (i-CAT, Imaging Sciences International Inc., Hatfield, PA) with a voxel size of 0.4 × 0.4 × 0.4 mm and a grey intensity of 12 bits. The CBCT scanning took 20 s, and, according to the manufacturer, had a radiation dose of 68 µSv at a tube current of 18.45 mAs and a tube potential of 120 kVp. This is similar to the dose previously measured for the maxillofacial region.²⁰ The segmentation and 3D reconstruction of the mandible and maxilla and radio-opaque markers from the CBCT data were performed using a commercial software system (Amira, Visage Imaging Inc., Berlin, Germany).

3D fluoroscopy

The 3D fluoroscopy method developed by Tsai *et al*¹⁹ for the knee joint was further developed in the current study to address the specific nature of the anatomy and movement of the mandible using a low-radiation exposure dental CBCT system, *i.e.* i-CAT, which was modified to enable fluoroscopy imaging.

Using the 3D fluoroscopy method, 3D skeletal kinematics were obtained by registering the 3D static CBCT bone models to the 2D dynamic fluoroscopic images (Figure 1). For this purpose, the formation of the fluoroscopic image of a bone was modelled as an ideal perspective projection of a point source X-ray through the bone onto the image plane. Each of the modelled X-rays went through a number of voxels of the CBCT volume of the bone in space. The Hounsfield numbers of these voxels were integrated along the ray and projected onto the imaging plane to obtain a digitally reconstructed radiograph (DRR) resembling a radiograph.^{21–23} A calibration object, consisting of a reseau plate and a star-shaped plate with lead markers at known positions was designed specifically for CBCT-based fluoroscopy for determining the position of the X-ray source relative to the image plane and to obtain parameters for image distortion correction.¹⁸ Since the CBCT and fluoroscopy were an integrated system, another calibration box with lead markers on specific positions was also designed and used to determine the relative poses of the coordinate systems for CBCT and fluoroscopy.

A CBCT bone model in space was defined as “registered” to the fluoroscopy image when its DRR (I_{DRR}) best matched the fluoroscopic image (I_{f}) according to a similarity measure called the weighted

Figure 1 Image processing steps for the registration of the mandible using the proposed weighted edge-matching score-based method. (a) A fluoroscopic image of the mandible (I_{fl}) with the radio-opaque markers marked in red; (b) a digitally reconstructed radiograph (DRR) by casting X-rays through the cone beam CT (CBCT) volume of the mandible (I_{DRR}); (c) dilation image (B_{fl}) of the fluoroscopic edge image E_{fl} with a small band; (d) the weighting image (L_{DRR}) obtained based on the length of the edges in the edge image of the DRR (E_{DRR}); (e) the registered CBCT models of the mandible (in purple) and part of the maxilla (temporal bone in blue) as seen from the X-ray source, giving DRRs which best match the fluoroscopy image; and (f) the registered CBCT models of the bones from an oblique view

edge-matching score (WEMS).¹⁹ For the calculation of the WEMS, the Canny method²⁴ was used to obtain edge images E_{fl} and E_{DRR} for I_{fl} and I_{DRR} , respectively. The image E_{fl} was then dilated to become B_{fl} with a dilation band of $(2d+1)$ pixels (Figure 1c). In B_{fl} , the value of a pixel within the band was set to be inversely proportional to its shortest distance from the original edge. In the current study, $d=2$ was chosen empirically. Similar to manual registration, long edges

in E_{DRR} were given greater weightings in the similarity score. The longer the edge to which one pixel belonged, the greater the weighting of that pixel. These weightings were stored in a weighting image L_{DRR} (Figure 1d). Each pixel of B_{fl} was multiplied by the value of the corresponding pixel of L_{DRR} to produce a registration image E_{reg} . The WEMS was then calculated as the total area of E_{reg} as a percentage of the total area of L_{DRR} as follows:

$$\text{WEMS}_{d,p} = \frac{\sum_{(x,y)} E_{\text{reg}}(x,y)}{\sum_{(x,y)} L_{\text{DRR}}(x,y)} \quad (1)$$

The greater the WEMS value, the closer E_{DRR} was matched with E_{fl} . A WEMS value of 0 indicated that no edges were matched, whereas the maximum WEMS value of 1 indicated that all edges were matched. Therefore, the registration of a bone model to the measured fluoroscopic image required finding the 3D pose (*i.e.* three translations and three rotations) of the bone model that minimized $\text{WEMS}_{d,p}$ [Equation (1)]. Since the problem was non-linear, a genetic algorithm was chosen to find the global minimum.²⁵ The genetic algorithm used a randomized initial population satisfying all bounds; rank-based fitness scaling; stochastic uniform selection of parents; scattered cross-over; and adaptive feasible mutation. The registration algorithm and the optimization were implemented with the Image Processing ToolboxTM and Global Optimization ToolboxTM in Matlab (The Mathworks, Inc., Natick, MA). All the analyses were performed on a personal computer with an Intel CPU (Core i7 860, 2.8 GHz), an NVIDIA graphical card (GeForce GTX260) and a RAM of 4 GB.

Experimental protocol

The necessary imaging for the *in vivo* measurement of the mandible kinematics was performed using the fluoroscopy function of the modified i-CAT system, giving images of sizes of 756×960 pixels with a pixel size of 0.254×0.254 mm at a sampling rate of 7.5 frames s^{-1} . The performance of the WEMS method was then evaluated *in vivo* based on the subject.

The subject, seated on a chair with her head fixed to the head support by a belt, was first imaged with fluoroscopy and CBCT (as described under Subject preparation) with the mandible in the rest position. Fluoroscopy imaging was also performed at static positions with the mouth open at distances of 1 and 2 cm, as well as during opening-closing and chewing at a self-selected pace for one complete cycle. For each test condition, the fluoroscopy imaging was performed for 10 s with an estimated radiation dose of 17 μSv at a tube current of 4.57 mAs and a tube potential of 120 kVp. Therefore, the total radiation dose of the experiment was about 153 μSv , about 15% of the limit of the acceptable annual dose (1 mSv year^{-1}) suggested by the United States Nuclear Regulatory Commission (USNRC).

Co-ordinate system definition

In the rest position, each bone was embedded with a technical co-ordinate system (TCS) defined by the radio-opaque markers. An anatomical co-ordinate system (ACS) of the mandible was also defined by the epicondyles and the centres of the edges of the two central incisors, which were digitized manually using a self-developed program. The mandibular ACS originated at the mid-point between the epicondyles, with the z -axis directed to

the right epicondyle, the y -axis directed superiorly and normal to the plane defined by the z -axis and the mid-point of the centres of the central incisor edges, and the x -axis as the cross-product of y -axis and z -axis and directed anteriorly. The ACS of the maxilla was chosen to be coincident with that of the mandible at the rest position. The poses of the bones were then described in terms of the three translations and three rotations of the ACS relative to the global co-ordinate system fixed to the image plane. Since the bones were considered to be rigid, the transformation between the TCS and the ACS for a bone was constant.

Error evaluation

The measurement errors were calculated as the differences in translations and rotations between the registered ACS poses using the WEMS method and those from the RSPA method.¹⁸ For each trial and each test, the bone models were registered to each fluoroscopic image using the WEMS method to give the poses of the registered ACS. The poses of the ACS using RSPA were obtained through the poses of the TCS defined by the relevant markers and the fixed TCS/ACS transformation. The measurement accuracy of the RSPA was evaluated using a porcine cadaveric model at 10 different bone poses by comparing the differences between 20 repeated RSPA measurements and the corresponding CBCT data. The movement of the markers on the maxilla during dynamic tests was also measured. A flowchart is given in Figure 2, depicting the procedure for the experiment, image registration and error evaluation.

Statistical analysis

The errors in the mandibular rotations and in the translations at both mandibular condyles were ensemble averaged over all the image frames of all the static tests (*i.e.* resting position and mouth openings of 1 cm and 2 cm) and over all the image frames of the movement cycle for each dynamic test (19 frames for opening/closing and 25 frames for chewing), giving means, standard deviations (SDs) and root mean squared errors (RMSEs). Mean values expressed the measurement bias, while the standard deviations indicated the precision.

Results

The RSPA method was found to have an accuracy of 0.37 (0.11) mm for the current marker positions. The maximum root mean squared movements of all the markers on the maxilla were less than 0.56 mm throughout the tests.

In static conditions, the means (SD) of the translation and rotation errors were all less than 0.1 ± 0.9 mm and $0.2^\circ \pm 0.6^\circ$, respectively (Table 1). The corresponding RMSEs were less than 0.9 mm and 0.6° (Table 1). For dynamic tests, the means (SD) of the translation and rotation errors over the movement cycle were

Figure 2 The general procedure of the experiments, image registration and error evaluation. ACS, anatomical co-ordinate system; TCS, technical co-ordinate system; WEMS, weighted edge-matching score

all less than 1.0 ± 1.4 mm and $0.2^\circ \pm 0.7^\circ$, respectively (Table 1). The corresponding RMSEs were less than 1.7 mm and 0.7° (Table 1). The patterns of the mandibular angles and translations relative to the maxillary co-ordinate system measured using the WEMS method were also very similar to those measured using RSPA during both open–close and chewing tests (Figures 3 and 4).

Discussion

The current study aimed to develop further a 2D-to-3D registration method for measuring 3D, *in vivo* kinematics of the mandible with single-plane fluoroscopy; to determine experimentally the accuracy and precision of the method; and to demonstrate its clinical applicability *via* measurements on a healthy subject during opening/closing and chewing movements. The proposed method was shown to have measurement errors less than 1.0 ± 1.4 mm for all translations and $0.2^\circ \pm 0.7^\circ$ for all rotations, much smaller than previously published errors associated with skin marker

movement artefacts using non-contact techniques.¹² These results suggest that the method can be used to measure the *in vivo* kinematics of the mandible during functional activities.

Methods for measuring 3D dynamic mandibular motion *in vivo* without the use of interfering transoral devices are expected to have great potential in relevant dental clinical applications. For patients without the necessary normal and stable teeth to fix the markers or transoral devices, some traditional measurement methods may become infeasible,^{10,11,26} especially for more complicated movements such as chewing. In contrast, the current method does not rely on markers or transoral devices, so it would be useful for accurate measurement of 3D, unlimited mandibular kinematics in the assessment of patients with missing or unstable teeth during chewing movements. Note that radio-opaque markers were necessary in the current study for the evaluation of the accuracy of the proposed markerless 3D fluoroscopy method. Since these markers might interfere with movement, the measured mandibular kinematics should be considered as part of the validation instead of a representation of the natural movement of the mandible. Further study using the 3D fluoroscopy method

Table 1 Means, standard deviations (SDs) and root mean squared errors (RMSEs) of the proposed method in measuring the mandibular kinematics during static tests, namely rest position, 1-cm and 2-cm mouth opening, and tests of opening/closing and chewing movements

Degree of freedom	Static			Opening/closing			Chewing		
	Mean	SD	RMSE	Mean	SD	RMSE	Mean	SD	RMSE
<i>x</i> (mm)	0.1	0.2	0.2	−0.1	0.2	0.2	−0.2	0.2	0.3
<i>y</i> (mm)	0.0	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1
<i>z</i> (mm)	0.1	0.9	0.9	0.2	1.2	1.2	1.0	1.4	1.7
α (deg)	0.1	0.5	0.5	0.2	0.5	0.5	−0.2	0.4	0.4
β (deg)	0.2	0.6	0.6	0.0	0.7	0.7	0.0	0.6	0.6
γ (deg)	0.2	0.3	0.3	0.0	0.3	0.3	−0.1	0.3	0.3

x, *y* and *z* are translations along, and α , β and γ are rotations about the anteroposterior, superoinferior and mediolateral axes of the maxillary coordinate system, respectively.

Deg, degree.

Figure 3 Mean curves of (top row) the rotations of the mandible about and (middle row, bottom row) translations of the most lateral projections of the mandibular condyles along the anteroposterior (A/P), superoinferior (S/I) and mediolateral (M/L) axes of the maxilla co-ordinate system during an open-close motion cycle. The curves measured by roentgen single-plane photogrammetric analysis are shown in grey while those measured using the weighted edge-matching score (WEMS) method without measurement bias are in black. The precisions of the WEMS method are also shown as error bars on the corresponding component curves

alone will be needed to provide accurate description of natural mandibular kinematics during activities.

The current method is based on single-plane fluoroscopy and also takes advantage of the CBCT system, enabling the measurement of the mandibular kinematics with a lower dose of radiation than bi-planar fluoroscopy approaches or standard medical CT systems. With the modified CBCT system, both CBCT and fluoroscopy imaging could be carried out without moving the subjects, largely reducing the time required for the measurement and any potential artefacts that might exist during transfer between systems and affect the subsequent data analysis. The proposed testing position with the maxilla stationary throughout the experimental process will also help reduce the effort in the registration of the maxilla in future clinical applications.

Experimental validation is critical in order to determine whether a measurement method is accurate enough for quantifying functional mandibular movement. However, few studies on the measurement of the mandibular motion have reported experimentally determined bias and precision of the measurement methods used.^{11,13} Since the accuracy of bone model-based 3D fluoroscopy

methods may be affected by the shapes, internal structures and motions of the target bones, it is necessary to evaluate experimentally the accuracy of the method for different joints in subsequent clinical applications.¹⁹ The current study was the first experimental validation of a 3D fluoroscopy method dedicated to the *in vivo* measurement of mandibular motion under real-life conditions. This was in contrast to previous validation studies of 3D fluoroscopy on other joints, which either used “clean” images of the bone outside the body²⁷ or only assessed the accuracy of the measurement at static conditions.²⁸ The current study reported the means and standard deviations of the measurement errors. Since the bias (mean) can be reduced by proper calibration (Figures 3 and 4), the precision (SD) becomes the key parameter for the performance of a measurement method. The proposed method was shown to have high precision in all motion components (Table 1). The precision of the out-of-plane translation was slightly less than for the other components as expected because with single-plane approaches the projected image of the bone model was less sensitive to out-of-plane translation than other components. Previous studies on other joints

Figure 4 Mean curves of (top row) the rotations of the mandible about and (middle row, bottom row) translations of the most lateral projections of the mandibular condyles along the anteroposterior (A/P), superoinferior (S/I) and mediolateral (M/L) axes of the maxilla co-ordinate system during chewing motion. The curves measured by roentgen single-plane photogrammetric analysis are shown in grey while those measured using the weighted edge-matching score (WEMS) method without measurement bias are in black. The precisions of the WEMS method are also shown as error bars on the corresponding component curves

have reported SDs of errors as large as 5.6 mm in the out-of-plane translation.^{28,29} With the WEMS method for the knee joint, Tsai *et al*¹⁹ reported precisions of less than 0.77 mm for in-plane translations, less than 1.13° for all rotations and 3.06 mm for the out-of-plane translation. The corresponding values for the TMJ in the current study were 0.2 mm, 0.7° and 1.4 mm (Table 1). These results indicate that the WEMS method, as applied to mandibular kinematics measurement, was more accurate than the WEMS method for measuring knee kinematics reported in Tsai *et al*,¹⁹ not only in translation components but also in rotational components.

There are several factors that may contribute to the better performance of the WEMS method in the current study than in the study by Tsai *et al*.¹⁹ As with any other bone model-based 3D fluoroscopy methods, the measurement accuracy of the current method may be affected not only by the similarity measure used but also by factors such as resolution of the CBCT data, pixel size of the fluoroscopic image, the projection geometry of the fluoroscopy system and the morphology of the bones. Since the CBCT and fluoroscopy systems as well as the targeted bones were all different in the

current study and the study by Tsai *et al*,¹⁹ without further systematic studies it is not clear which specific factor would have the greatest effects on the differences in the performance. Nonetheless, the proposed method with the modified CBCT/fluoroscopy system can be considered an accurate method for measuring 3D mandibular kinematics *in vivo* during both static and dynamic activities.

In conclusion, a 3D fluoroscopy method based on an existing similarity measure was proposed for measuring *in vivo* the kinematics of the mandible during static and dynamic movements using single-plane fluoroscopy. The method was shown experimentally to have measurement errors less than 1.0 ± 1.4 mm for all translations and $0.2^\circ \pm 0.7^\circ$ for all rotations. Without relying on transoral devices, the proposed method will be useful for *in vivo* research and clinical measurements of 3D kinematics of the mandible during functional activities.

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