TECHNICAL REPORT

Automatic fusion of lateral cephalograms and digital volume tomography data—perspective for combining two modalities in the future

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Objectives: This article investigates the combination of three-dimensional (3D) digital volume tomography data with two-dimensional (2D) cephalograms in dentomaxillofacial imaging.

Methods: An automatic hierarchical method to adjust the geometrical relations of these two modalities is presented. The approach is tested on phantom and patient case data, where the feasibility, usability and potential possibilities of the presented innovative method are highlighted. Digitally reconstructed radiographs are computed by casting rays through the 3D volume to get a 2D projection of the volume to produce realistic simulated cephalograms. Different similarity measures are considered based on variations of statistical and deterministic optimization procedures. Stability, precision and accuracy of the method are investigated.

Results: The presented algorithm demonstrates a reasonable solution of the corresponding 2D/3D registration problem. Exemplary results from phantom and patient case data are presented. Tooth movement could be determined, in contrast to the 2D lateral cephalogram, separated for each side in all three spatial directions.

Conclusions: Achieved results are highlighted from a clinical point of view and demonstrate the clinical benefit in daily praxis.

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Introduction

Two-dimensional (2D) lateral cephalograms (CEPHs) were introduced into orthodontics in 1931 by Broadbent.^{1,2}

The cephalometric examination became an inherent part of orthodontic diagnostics and treatment planning.³ Standard cephalometric radiographs are inherent 2D records of a three-dimensional (3D) scene. 3D diagnostics for dentistry based on CBCT was presented for the first time in 1998 by Mozzo et al.⁴ CBCT is

a medical imaging technique that generates 3D images at a lower cost and absorbed dose than conventional CT.⁵ Today CBCT, often termed as digital volume tomography (DVT) in dentistry, is used in different disciplines of dental medicine and is common in orthodontics.⁶ The superimposition of 2D CEPH and 3D DVT data could open various new diagnostic capabilities. Orthodontists, for example, could measure anatomical structures for treatment planning in a more accurate way than is possible today. It is conceivable to assign the volumetric jaw data three dimensionally to

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cephalometric landmarks. For example, in cephalometric radiographs (2D), this is commonly at the anterior base of the skull.

In order to combine CEPH and DVT data, the image acquisition geometry of the two modalities must be aligned to a common measurable mathematical basis. This problem is in general called image registration. A variety of different image registration techniques were presented by Brown,⁷ and Maintz and Viergever.⁸ The technical focus in this work will be 2D/3D image registration, which is necessary to align two different acquisition geometries and co-ordinate systems and is typically known from image-guided interventions.⁹ Beyond 2D/3D image registration, methods are also often used for X-ray and fluoroscopy or 3D-CBCT and 2D-CT modalities in image-guided radiation therapy.^{10,11} Today, there is no comparable work on the field of X-ray-based multimodal image registration in dental imaging.

Currently, data fusion from both modalities takes place only in the mind of an observer. Therefore, the problem discussed in this article consists of an automatic computer-aided combination of the two modalities DVT and CEPH in one common context. A new framework for representing 2D and 3D data also in terms of visualization and data fusion is introduced and may open completely new diagnostic possibilities.

The main aim of this article is a feasibility analysis of automatic registration of lateral CEPH and DVT data. Therefore, digital reconstructed radiographs (DRRs) are calculated on the basis of volume tomography data to allow a comparison in a 2D subset of the data space. The goal is to identify transformation parameters, which lead to a projection of the volume that is as similar as possible when being compared with the lateral CEPH. For this purpose, a variety of similarity measures and optimization algorithms are known from literature.¹² Since for dental X-ray-based imaging no techniques are known from literature yet, in this work, a new hybrid approach based on a combination of statistical similarity measures and optimization procedures will be presented. The feasibility of the implemented algorithm is verified on the basis of a selection of phantom and patient data. The results will be highlighted from a clinical point of view.

Methods and materials

Data acquisition

All 2D and 3D data sets were acquired using one DVT machine (Orthophos[®] XG 3D; Sirona Dental Systems GmbH, Bensheim, Germany). The phantom head was mounted on a steady tripod for cephalogram and DVT exposure. For phantom and patient acquisition, the machine was operated with 8 × 8-cm field of view settings. For this study, three real patient registration cases who underwent DVT examination for medical reasons as well as four phantom data cases have been considered for feasibility analysis and evaluation of the proposed methods.

Data handling and hardware

We used 16-bit raw CEPH images as data input with a resolution of 1804 × 2136 pixels and 0.104 mm pixel size. For reasons of performance and robustness, we reduced the resolution of these images in all phases by a factor of 4. The reduction of the image size is comparable to a smoothing filter and raises at the same time the signal-to-noise ratio. The second input, the DVT volume, has a resolution of 512 × 512 × 512 voxels by a voxel size of 0.16 mm. The algorithm was developed and tested on a Intel[®] Quad Core CPU (Intel, Santa Clara, CA) with 2.1 GHz, 8 GB memory and GeForce[®] Kepler graphics adapter with 2-GB memory. DRR generation was implemented with NVIDIA[®] CUDA[™] v. 3.0 (NVIDIA, Santa Clara, CA).

Geometrical set-up and co-ordinate system

In X-ray imaging, positioning of an object of interest in the machine usually suffers a huge scan-to-scan variability. Additionally, the machine dynamics, especially the gantry motion—defined by the position of the source and detector over time—is also not constant for all acquisitions. These variabilities can be expressed by a rigid body motion M , which is mathematically represented by a rotational matrix R and translational vector T , as given by Equation (1).

$$M = [R|T] \quad (1)$$

This transformation formalism is known as the extrinsic part of an imaging system.¹³ While in 3D imaging, it is crucial to know M for any gantry situation, in 2D acquisitions, *e.g.* CEPH, M is usually not calibrated and unknown. In order to describe an imaging system in addition to the extrinsic parameters, the attributes of the camera system is given by an intrinsic camera matrix A [Equation (2)], with source to detector distance f , detector pixel spacing in both detector directions α_u and α_v , and possible shift of the detector (u_0, v_0).

$$A = \begin{bmatrix} \frac{u_0}{f} & \alpha_u & 0 \\ \frac{v_0}{f} & 0 & \alpha_v \\ \frac{1}{f} & 0 & 0 \end{bmatrix} \quad (2)$$

The combination of extrinsic and intrinsic parameters can be represented by the projection matrix P , as given by Equation (3).

$$P = A \cdot M = [AR|AT] \quad (3)$$

P represents a homogeneous 3 × 4 matrix that maps a 3D position in space to a 2D plane of the imaging system (detector) and thus describes very well the fundamental geometrical situations of an X-ray projection,

Figure 1 A cephalogram is a perspective X-ray projection of the patient on a detector. This geometry can be used as a model for creating a virtual cephalogram of a digital volume tomography volume by mapping points in a three-dimensional co-ordinate system to a two-dimensional detector.

including gantry and detector position and orientation in a 3D world co-ordinate system.

We use this formalism to define a perspective projection of a representation of a patient given by a DVT volume in a cephalogram X-ray set-up (Figure 1). The co-ordinate system is defined with its origin in rotation centre. In addition, the 3D volume has its centre here too. The transformation M in Figure 1 illustrates the extrinsic parameters that describe the six degrees of freedom (DOF) of patient positioning. The intrinsic transition described by A defines the mapping onto the detector plane. The y - and z -axis are parallel to the u - and v -axis (rows and columns) of the 2D detector plane. We define the x -axis parallel to the normal of the detector plane. In the following, we considered the ideal Feldkamp case with an ideal detector and no aberration beyond line of sight. Since the intrinsic parameters (f, a_u, a_v, u_0, v_0) of the set-up can be assumed as constant and well known for a real X-ray imaging machine, only the position and orientation of the DVT volume with respect to the cephalogram, which represents six DOFs are unknown. In the following, these unknown parameters are referenced by $s = (t_x, t_y, t_z, \varphi_x, \varphi_y, \varphi_z)$ while $T = (t_x, t_y, t_z)$ is the translation, $(\varphi_x, \varphi_y, \varphi_z)$ are the Euler angles leading to the rotation matrix R . All calculations are based on rigid body movements, and non-rigid body movements are not considered.

Digital reconstructed radiograph—virtual projection of a three-dimensional volume

We used a 3D DVT volume as a representation of a real patient and used Equation (3) to create a virtual CEPH image on a virtual detector plane. By this, the virtual projection, known as DRR, could be compared to a real X-ray CEPH image of the patient in a 2D subspace.

DRRs can be generated by several algorithms, such as ray-driven and voxel-driven approaches. We decided to

use the ray-casting method, because this treatment is closer to the real physical process of the X-ray imaging technique, is less sensitive to interpolation artefacts and can be parallelized efficiently on graphical processing units.

A typical situation of the ray-tracing technique for a solution $s(i)$ (one iteration i) is depicted in Figure 2. As one can see, for each pixel on the detector plane, at least one ray is simulated from a source point to the detector pixel. Pixel and source positions can be determined very quickly from the compact representations of the projection matrix.¹³ All voxels that are hit by a virtual ray are accumulated to get pixel values on the receiving detector plane. Interpolation over surrounding voxels is made trilinear on graphical processing units. As one can see, for the registration of CEPH and DVT data, the volume is quite close to the detector as the real patient is in the case of a real CEPH.

Optimization and similarity measurement

Since a DRR generated from a patient's DVT—as described before—can be compared with a real patient's CEPH image in a common solution space, one can formulate an optimization problem to find the best solution s , with regard to a measureable error *e.g.* given by a L2 norm $\| \cdot \|$, as described by Equation (4).

$$\| \text{CEPH} - \text{DRR}(s) \|_s \rightarrow 0 \quad (4)$$

Since especially for DVT data DRRs usually will not fully match to a real X-ray image, *e.g.* owing to truncation and cone beam artefacts, even if the solution s is optimal, one usually considers similarity measures instead of magnitudes or norms and tries to maximize the similarity of CEPH and DRRs. A comparison of similarity measures (metrics) for use in 2D/3D medical image registration can be found in Penney *et al.*¹² In our study, three metrics show appropriate convergence capabilities: Mutual Information, especially Mattes Mutual Information, is the first choice for similarity measurement of multimodal images. Additionally normalized cross correlation and a gradient-based correlation are considered as well. More details on Mattes Mutual Information can be found in Mattes *et al.*¹⁴ and Mattes *et al.*¹⁵ Normalized cross correlation is a correlation-based metric and proves the linear coherence of the two images. Additionally, we used normalized cross correlation based on gradient images. For this purpose, both images were previously high-pass filtered, so that only edges of the bone structure remained.

The choice of the optimizer has an important impact on how the solution space is sampled and thus on the convergence quality and velocity of the algorithm. For our study, we introduced a novel-nested optimization approach, which is sketched in Figure 3. We could show that by this strategy, we could solve the optimization problem (Equation (4)) in an acceptable time and in a stable manner.

The new evaluated nested registration algorithm can be subdivided into three phases. Each phase includes an

Figure 2 The virtual X-ray projection of a digital volume tomography volume for a cephalogram set-up. The distance between virtual source and detector is, proportional to the size of DVT, large. This results in an almost parallel projection geometry. Since here a small digital volume tomography volume is used, most of the virtual image will be empty, because no voxels are crossed by any virtual rays.

iterative optimization procedure. The result of each phase is used as the initial value in the corresponding successor phase. Figure 3 shows a flowchart of the three registration steps. With increasing levels the number of optimization parameter, iteration and the quality of the registration result is rising. On the other hand, the search radius, the region of interest and the convergence criteria, and the value where the optimization algorithm stops are decreasing. The iterative part of our presented algorithm is highlighted in this figure as a separated block, which shows the essential method for 2D/3D registration.

Phase 1: In the first phase of the algorithm, a rough registration with only two translation parameters in 2

DOF (t_y, t_z) is considered to stabilize the search region of the solution space. The two parameters correspond to a movement parallel to the detector plane. All other transformation parameters remain at its default value. This phase leads at the end to a rough registration of the DRR and cephalogram image.

Phase 2: The DOF, that is the number of optimization parameters, increases in this phase to six (three for translation, and three for rotation). For optimization of these parameters a probabilistic search algorithm is used prohibiting the algorithm to get trapped in local minima. The step size and search radius is still kept wide. The result of this phase has to be interpreted as a rough registration.

Figure 3 Overview of the introduced nested registration algorithm with an iterative two dimensional (2D)/three dimensional (3D) registration part for each phase. Essentially, the results of each phase work as inputs for successors, while the degrees of freedom increases and the sensitivity to solution variability decreases. S_0 determines the initial transformation parameters, S_1 are updated parameters during the procedure and S_{END} are the final parameters. CEPH, cephalogram; DRR, digitally reconstructed radiograph; DVT, digital volume tomography.

Phase 3: The convergence criterion is formulated to be much more restrictive for this phase, which leads to an increasing computation cost in comparison to prior phases. As a consequence, the number of search iterations will rise, while the quality of the results during these iterations will increase.

Results

It turns out that for all input data, the presented hierarchical registration function superior a common one step registration algorithm. This hierarchical solution reduces the actual required change in geometry step by step, which is needed to improve the superimposition, during each phase.

The non-trivial task of the given registration algorithm is to find a minimum in a very flat area although it is the only minimum in the surrounding area. Therefore, one main challenge is to stabilize the solution by the choice of a tolerant optimizer. Figure 4 shows an extraction of the solution space for translation in the (t_y , t_z) direction. Small values represent a better similarity measure. Our approach with an evolutionary optimizer emphasizes the most robust solution compared with the pervasive simplex and Powell method, especially for the first phases.

By increasing iterations and time, the measured similarity value converges against the global minimum, as shown in Figure 5. The convergence map illustrates that at the beginning of each registration phase the variation of the measured similarity function is kept very high to

avoid convergence in local optima, which can be compared with stabilizing by a simulation annealing technique.¹⁶ The three different registration phases presented in this article can be clearly seen in Figure 5, indicated by P1, P2 and P3. The similarity measure for the 3D volume and 2D image at the end of our registration algorithm reaches after $i = 1200$ iterations exactly the minimum that was determined by a brute force method.

Each iteration i requires 44 ms, which includes 19 ms for DRR generating, leading to a full computational time with an average of 60 s including input and output interactions with the file system.

The quality of the optimizer can be determined by a comparison of the result with a full sampling of the solution space, here defined as brute force method.

Figure 6a–f demonstrates a 3D view of combined CEPH and DVT data, using the result of the presented method on phantom data. The superposition of the cephalogram and DVT volume is true to scale and shown from different angles. The conformity of the cephalogram and DVT image is excellent particularly in the anterior teeth area. The observer obtains a fully new impression of how to display the cephalogram image and the DVT volume. Both can now be shown simultaneously in the same view. Since the CEPH is a ray integration of the DVT voxels, it can not only be seen as a representation of one DVT layer. But since the ray and cone geometry is calculated it can be shifted true-to-scale by the observer towards the source or detector direction to clearly evaluate which corresponding volume structures contribute the final CEPH image.

Figure 4 Two-dimensional extraction of the solution space for a real patient case optimization. While the map shows the t_y and t_z sensitivity and convex behaviour of the optimization criteria for these two variables, the other parameters are close to optimum.

Figure 5 An optimization convergence map for a real patient case. As a similarity measure the mutual information is considered. A brute force search indicated the optimal solution at the bottom of the graph. One can see the different phases of the presented hierarchical optimization, which show the convergence towards the optimal (brute force) solution.

Figure 7 shows phantom data with a 3D measurement line between 16 (World Dental Federation two-digit notation) palatal and the sella turcica planes, which is positioned perpendicular to the nasion–sella

turcica conjunction. As one can see, by our method, a measurement independent to the lateral view is possible. Thus for 26 palatal to the sella turcica plane, a different and more accurate measurement is possible.

Figure 6 Different views (a–f) of three-dimensional rendering of a phantom, superimposed with lateral cephalogram based on hierarchical two dimensional/three dimensional registration algorithm. Both cephalogram and digital volume tomography were acquired with Orthophos[®] XG3D (Sirona Dental Systems GmbH, Bensheim, Germany).

Figure 7 Three-dimensional (3D) rendering of a phantom, superimposed with lateral cephalogram based on hierarchical 2D/3D registration algorithm. Additionally, an example 3D measurement line between 16 palatal and the sella turcica planes, which is positioned perpendicular to the nasion–sella turcica conjunction. Both cephalogram and digital volume tomography were acquired with Orthophos® XG3D (Sirona Dental Systems GmbH, Bensheim, Germany).

Once the geometrical relations are known, one is also capable of combining the final resulting DRR with the CEPH image in a 2D space. This superposition is depicted in [Figure 8](#) for a skull phantom and in [Figure 9](#) for a real patient case exemplarily. Patient cephalogram images usually indicate a different jaw position between upper and lower jaw for multiple acquisitions. For that reason, a segmentation of the upper jaw is realized before registration. Afterwards only the upper jaw was used as input for the registration algorithm.

Discussion

Today, there is no comparable work in the field of X-ray-based multimodal image registration especially for dental imaging known from literature or state of the art. Therefore, this work has mainly the character of feasibility demonstration and certainly further investigation regarding clinical and technical aspects will follow in future work. Our results demonstrate the applicability of our method for general practical usage. To analyse the quality of proposed methods a target/actual performance comparison would certainly be the best representation to quantify the optimization error, which was not possible during this study. However, inaccuracy of controlling a dedicated positioning system will be in the order of submillimetres and thus induce a systematic error too. The perceptual registration quality was very promising for phantom as well for patient cases. Since no systematic clinical evaluation has been considered by this work yet, future work should point out expected registration quality for relevant diagnostic and clinical purposes.

To provide a quantitative measure, the authors decided to measure the algorithmic limit of accuracy of the proposed method itself, which can be seen as a natural

Figure 8 Superimposition of lateral cephalogram (grayscale) and digital volume tomography (DVT) (coloured) phantom data after hierarchical nested three-step registration algorithm. Bright spots signalize high coverage, both in anterior teeth and back tooth area. Slightly raised noise in the gingiva area is visible. The field of view (DVT) is isotrop 8 cm. For colour image see online.

Figure 9 Superimposition of lateral cephalogram (grayscale) and maxilla digital volume tomography (coloured) from patient data after hierarchical nested three-step registration algorithm. Bright spots signal high coverage, both in anterior teeth and back tooth area. Slightly raised noise in the gingiva area is visible. The field of view (digital volume tomography) is isotrop 8 cm. For colour image see online.

limit of registration resolution. The investigation was based on the global behaviour of the similarity measure in the solution space close to the global optimum, as depicted in Figure 10.

To verify the accuracy of our results, we investigated the solution space around the global minimum more precisely. This is depicted for one DOF, here (t_y), in Figure 10. One can see a systematically full search scan of the similarity measure's solution space, which is very noisy in this area and therefore the quality of the optima is determined by the amount of noise, which may be represented by the standard deviation σ . The measured data $\hat{F}$ are fitted to a polynomial curve F with the degree of four. Then $R = \hat{F} - F$ defines a noise estimate of the similarity measure. For 3σ , 99% of the various values of R lie inside the symmetric interval around the expected value. This border $|F_{(0)} - \hat{F}_{(0+e)}| \leq 3\sigma$ is considered as the maximal resolution accuracy, leading to $e \leq 0.3$ mm.

Owing to the magnitude of e , which is close to the resolution of DVT volume (0.2 mm) and the periodic characteristics of the noise, we assumed that the limitations of this method were closely related to the sampling of the volume space. One has to investigate whether the resolution of this 2D/3D problem can be

higher than the voxel limitation of the input DVT volume in future research.

The radiation dose for this kind of radiograph adds up to $30 \mu\text{Sv}$, depending on the selected programme. This is comparable with two digital orthopantomogram or lateral CEPH acquisitions with $15 \mu\text{Sv}$ each.^{17,18} For modern commercial DVT units a range of 11 – $674 \mu\text{Sv}$ is denoted by Ludlow,¹⁹ where $30 \mu\text{Sv}$ is located in low range.

This article outlines that the presented hierarchical 2D/3D registration algorithm is capable of combining lateral CEPH and 3D DVT data. The observer gets a completely new opportunity for visualization of these modalities. Especially for orthodontists this superimposition could be helpful.

The possibility of a 3D visualization of the jaw for determining cephalometric lines or planes allows a quantitative analysis of therapy-induced tooth movement.

Tooth movement could be determined, in contrast to the 2D lateral cephalogram, separated for each side in all three spatial directions Figures 6a–f and Figure 7. As already mentioned the radiation dose is higher and it is not advisable to use this kind of acquisition routinely for every patient, it should be reserved to chosen orthodontic patients with strict medical indication. Nevertheless, the novel method demonstrates potential benefits for existing data fusion, *e.g.* for pre- and post-interventional comparison. Further investigation, especially in combination with low dose imaging programs are necessary.

Conclusions

Our hierarchical 2D/3D registration algorithm, specifically developed for dental imaging, indicates a feasibility study based on three real patient datasets who underwent DVT examination for medical reasons as well as for four phantom datasets. The assessment of our results is focused on the superposition of the two different modalities that have been considered individually by the observer so far.

Although robustness was not systematically analysed in this work, our approach for the automatic superposition of CEPH and DVT data demonstrates convergence in reasonable time without any user interaction prior to the registration phases for all considered test cases. The authors show that registration results may be visualized in different ways, enabling new possibilities for measurement and diagnostic assessment for the observer of the data. The authors believe that the demonstrated feasibility study has the potential from technical and clinical point of views to find the way from bench to bedside and can create and add value in dentomaxillofacial imaging. The observer finally gets a fully automatically calculated image with the superimposition of 2D CEPH and 3D DVT data.

Figure 10 One-dimensional extraction of solution space around the global minimum with measured and fitted curve data.

Nevertheless, owing to higher radiation dose for this method it should be reserved to chosen orthodontic patients or for cases where data from both modalities

are available. The integration and further evaluation of this technique for clinical usage should be part of future work.

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